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# Rearfoot, mid/forefoot, and barefoot running: biomechanical differences related to injury

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**Rearfoot, mid/forefoot, and barefoot running:  
Biomechanical differences related to injury**

by

**Elizabeth Rose Boyer**

A dissertation submitted to the graduate faculty  
in partial fulfillment of the requirements for the degree of

**DOCTOR OF PHILOSOPHY**

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2015

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**ABBREVIATIONS**

AP	anterior-posterior
BF	barefoot
BW	body weight
COM	center of mass
CT	computed tomography
EMG	electromyography
FFS	forefoot strike
GRF	ground reaction force
hFF	habitual mid-/forefoot striker
hRF	habitual rearfoot striker
ITB	iliotibial band
LL	leg length
LR	loading rate
MFS	midfoot strike
ML	medial-lateral
PSL	preferred stride length
RFS	rearfoot strike
SI	strike index
SL	stride length
SW	step width
VIP	vertical impact peak
VLR	vertical loading rate
VO <sub>2</sub>	volume of oxygen/oxygen consumption

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## ABSTRACT

Running is an accessible, inexpensive form of exercise. However, running injuries are common and burdensome. Many runners are curious about switching to barefoot (BF) running or minimalist shoes to avoid injuries, which is commonly associated with rearfoot strikers (runners who land on the posterior third of their foot) landing on the middle or front third (mid- or forefoot strike; FFS). My purposes were to determine if undocumented aspects of FFS and BF running are beneficial or harmful compared to rearfoot striking (RFS), and if any potential benefits of FFS or BF running can be achieved by shortening one's stride length (SL) when using a RFS. I addressed this by conducting four studies. For all studies, habitual rearfoot (hRF) and habitual mid/forefoot strikers (hFF) ran with a RFS, FFS, and BF, and with a shorter SL. I looked at several ground reaction force (GRF) variables, kinematics, joint kinetics, tibial accelerations, stride time variability, and tibial bone stresses.

Impact peaks were present in the vertical GRF for RFS and in the posterior and medial directions with a FFS. Loading rates were generally higher in those same respective directions for the two foot strike styles. hRF decreased their vertical GRF variables by using a FFS.

Joint moments and contact forces were generally larger at the ankle for FFS while they were similar or larger at the hip for RFS. Some knee moments and contact forces were larger for RFS while others were larger for FFS. Patellofemoral loads were similar for foot strike styles. Nearly all moments and contact forces decreased with a shorter SL.

Free moment was distinctly different between foot strikes. Additionally, step width was narrower for FFS, but ITB strain and strain rate were similar, while RFS demonstrated greater pelvic drop. Shortening one's SL had many minor beneficial effects, including a wider SW, and decreased free moment, pelvic drop, hip adduction, ankle eversion, ITB strain and strain rate.

Stride interval long range correlations became more patterned for the novel conditions, suggesting runners were not operating in their optimal state of adaptability, possibly making them susceptible to injury.



Finally, FFS resulted in higher compressive, tensile, and shear stresses in the distal tibia compared to RFS and BF. An 8% shorter SL only significantly decreased shear stresses.

In conclusion, not all lower extremity loading variables are lower with a FFS or when running BF; in fact, several variables were elevated compared to RFS. Therefore, switching to a FFS (whether shod or BF), may not alleviate certain pain or help runners avoid injury. A more viable option may be to shorten one's SL 10%.

## CHAPTER 1

### INTRODUCTION

#### **Dissertation Organization**

This dissertation is a collection of four studies (five manuscripts), preceded by a general introduction. All studies are all related to the current, highly-debatable topic of running with or without shoes and the two major foot strike styles—rearfoot strike or non-rearfoot strike (midfoot or forefoot strike). My purpose is not to take a side on this issue but rather to present objective findings from studies conducted in our lab. I hope these data will help runners make an educated decision if they are contemplating switching footwear and/or foot strike styles, and ultimately help decrease the occurrence of overuse running injuries.

#### **General Introduction**

Recently barefoot running (BF) or running in minimalist shoes has become popular, and usually accompanying this is a switch from initially landing on one's heel (rearfoot strike) to landing on the middle or front third of the foot (midfoot/forefoot strike). Promoters of switching running styles claim it may reduce risk of injury. Some research supports this, as the vertical ground reaction force (GRF) impact peak, vertical loading rates (VLR), and knee loads are lower with a mid/forefoot (FFS) versus rearfoot strike (RFS). There have been a couple small retraining studies where habitual rearfoot strikers (hRF) with patellofemoral pain or anterior compartment syndrome switched from a RFS to FFS, resulting in reduced pain, symptoms, vertical ground reaction force impact peak, and loading rates. A few other retrospective studies noted greater incidence of injury in shod runners and/or hRF as compared to hFF and BF runners. However, there are no large prospective studies objectively comparing overuse injury rates between hRF, hFF, or BF running. One case study showed two men switching to a FFS in barefoot simulating shoes both sustained metatarsal stress fractures, and another study found 10 out of 19 runners had increased bone edema or fractures in their foot after training in minimalist shoes. Similarly, preliminary data from our lab and another lab showed some tibial bone stresses and strain rates are greater in shod FFS versus shod RFS running. Further research in this area is vital. Several other studies have

shown that shortening runners' stride length (SL) may decrease impact peaks, loading rates, joint work, knee loading, and risk for tibial stress fractures in hRF. If this is the case, we may help runners avoid unnecessary injuries by having them shorten their SL rather than switching foot strike styles.

Collectively, I hope to address if some undiscovered aspects of FFS and/or barefoot running are beneficial or harmful compared to RFS, and if any of these potential benefits of FFS or BF can be achieved by shortening one's SL while maintaining a RFS.

## **CHAPTER 2**

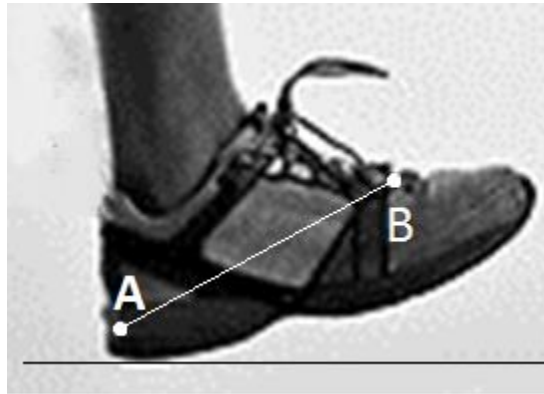
### **LITERATURE REVIEW**

An estimated 29 million Americans run  $\geq 50$  days per year (2013 State of the Sport), but annual injury rates range from 19-79% (van Gent et al., 2007). Running is also an essential component of military training, which encompasses another large population of persons sustaining overuse injuries related to running (Knapik et al., 2001). Injuries to the shank usually rank as the second or third most common running injury, only behind knee injuries (Taunton et al., 2002; 2003). Stress fractures account for approximately 1-20% of these injuries (Fredericson et al., 2006; Snyder et al., 2006). The tibia is the most common location of these stress fractures in runners, accounting for 33-50% of all stress fractures (Matheson et al., 1987; McBryde, 1985), while it is usually the second most common location in military personnel, after the foot (Jones et al., 2002; Pester & Smith, 1992; Zadpoor & Nikooyan, 2011). Even though knee injuries are the most common (Taunton et al., 2002; 2003), stress fractures are among the most serious because they require four weeks or more of rest (Zadpoor & Nikooyan, 2011). My series of studies will indirectly help identify runners at greater risk of sustaining overuse injuries, primarily bony injuries, and what changes to their running may decrease risk of injury. Because of the commonality of tibial stress fractures in runners, I will primarily be focusing on loads in the lower leg.

#### **Defining foot strike styles**

In general, researchers classify foot strike during running by where initial contact is made with the ground along the length of the foot. If available, foot strike style is usually classified using a combination of ground reaction force (GRF) data from a force platform and kinematic data, or marker positions, which gives us a strike index (SI) (Cavanagh and LaFortune, 1980). SI is calculated as the initial location of the GRF vector expressed as a decimal or a percent of foot length. Contact in the posterior third is a rearfoot strike (RFS, SI  $< 33.3\%$ ), middle third is a midfoot strike (MFS, SI: 33.3-66.6%), and anterior third is a forefoot strike (FFS, SI  $> 66.6\%$ ). Kinetics are not always available, so kinematic cutoffs have been proposed (Altman & Davis, 2012a). They defined the foot as a straight line from

the heel marker to a marker on the dorsifoot (see Fig. 1), and considered standing foot strike angle to equal  $0^\circ$  (i.e., this angle was subtracted from all foot strike angles during running trials). Their classification of foot strike angles during shod and barefoot (BF) running is in Table 1. Overall, when shod, classification based solely on kinematic data was sufficient (82% correct,  $r^2=0.85$ ) and better than BF running classification (60% correct,  $r^2=0.74$ ). Objective assessment of foot strike style using kinetics and kinematics is preferred over just kinematics, which is preferred over self-report, in which approximately one-third of runners misjudge their foot strike style (Goss et al., 2012; Goss et al., 2015).



**Figure 1.** Foot strike angle is angle AB forms relative to the ground, after subtracting angle AB during standing (*adapted from Altman & Davis, 2012*).

**Table 1.** Foot strike angle ranges ( $^\circ$ ) for the 3 foot strike categories averaged across shod and barefoot, and correct classification based on the strike index (SI) gold standard.

	RFS	MFS	FFS
<b>Foot strike angle (<math>^\circ</math>)</b>	$>8.0$	$-1.6$ to $8.0$	$< -1.6$
<i>Correct classification compared to SI</i>			
<b>Shod</b>	28/29 (96.6%)	5/6 (83.3%)	16/25 (64%)
<b>Barefoot</b>	0/2 (0%)	1/3 (33.3%)	11/15 (73.3%)

RFS: rearfoot strike, MFS: midfoot strike, FFS: forefoot strike

### Prevalence of foot strike styles

Considering shod runners, approximately 75-92% are habitual heel or rearfoot strikers (hRF), and 8-25% are non-heel strikers (3.4-23.7% MFS and 1.4-1.8% FFS) (Hasegawa et al., 2007; Larson et al., 2011; Samaan et al., 2014). Some runners (5.9%) may use different foot strikes between feet, and the prevalence of RFS may increase later in a run (Larson et al., 2011) and at slower speeds (Breine et al., 2014). These estimates, however,

may not truly represent recreational runners during normal training, because Hasegawa et al. (2007) looked at international elite runners during a half-marathon at the 15 km point, and Larson et al. (2011) measured recreational or sub-elite athletes at the 10 km point of a half marathon. Considering novice runners, approximately 97% of people wearing a neutral running shoe used a RFS, less than 2% used a MFS/FFS, and less than 2% used asymmetrical strike patterns (Bertelsen et al., 2013). Therefore, the true prevalence of shod runners who are hRF is likely between 75-97%.

### **Why do people switch foot strike and/or footwear?**

#### Injury prevention

The advent of minimalist shoes within the last 10 years and interest in barefoot running has recently caused much hype in the running community. Additionally, *Born to Run*, a book published in 2009 that uses objective and subjective data to encourage running barefoot or in minimalist shoes, has also spawned such interest. In a 2012 survey of 785 runners, over 75% of runners said they were interested in running barefoot (BF) or in minimalist shoes, primarily to reduce their chance for injury in the future (Rothschild, 2012), but 54% of them were still hesitant to try minimalist shoes or BF in fear of a new injury. Goss & Gross (2012) found similar trends in a survey of 2509 runners, as ~45% of runners who switched footwear or foot strike patterns (most from being hRF in traditional shoes) did so because of injury. A similar 2014 study of 509 runners discovered over half of respondents were partaking in some BF running because of the media hype and to decrease injury risk (Hryvniak et al., 2014). And since anywhere from 1-4 in 5 runners sustain an injury annually (van Gent et al., 2007), runners will try something new to avoid injury.

However, there is conflicting evidence of absolute injury reduction with a FFS and/or BF running or barefoot-simulating footwear (e.g., Vibram FiveFingers), as some runners endured new injuries while others report injuries disappearing (Cheung and Davis, 2011; Daoud et al., 2012; Diebal et al., 2012; Giuliani et al., 2011; Hryvniak et al., 2014; Ridge et al., 2013).

#### RFS vs. FFS

Two studies (n=3 and 10) reported all shod hRF who converted to a FFS relieved their patellofemoral pain or anterior shin pain (Cheung and Davis, 2011; Diebal et al., 2012), and anecdotally, the Spalding Running Clinic at Harvard University (headed by Dr. Irene

Davis) has noted many participants alleviating their running-related pain by switching to a FFS and/or BF. A retrospective study has found about 2-times higher injury rates in 36 hRF than 18 hFF (Daoud et al., 2012), although this association was minimized and bordered on insignificance after removing runners who alternated between foot strike styles (n=9) (likely because of lack of statistical power). Approximately 84% of these runners sustained some gradient of injury, so with such a small sample size and classification ambiguity, conclusions about injury rates for different injuries between foot strike groups are tenuous. Goss & Gross (2012) surveyed over 900 runners and found that 52% of self-identified hRF reported injuries in the last year while only 35% and 23% of midfoot and forefoot strikers experienced an injury, respectively. However, approximately one-third runners misclassify their foot strike style (Goss et al., 2012; Goss et al., 2015), so the validity of self-reported foot strike style injury rates is questionable, especially since <50% of those surveyed in Goss & Gross (2012) reported being hRF (empirically, ~75-97% are hRF (Hasegawa et al., 2007; Larson et al., 2011; Bertelsen et al., 2013)).

#### *Shod vs. BF*

Hryvniak et al. (2014) surveyed 509 runners and reported that 69% of runners starting barefoot or minimalist shoe running had previous injuries abate, and 64% did not report any new injuries (Hryvniak et al., 2014). The most common injuries that improved after incorporation of BF running included: knee (46%), foot (19%), ankle (17%), hip (14%), and low back (14%). As expected, a majority of runners suffered foot or triceps surae related pain (55%) when initially transitioning to barefoot running. Naturally, the foot and ankle accounted for 1/3 of the new injuries in runners who reported injuries (36%). In over 900 runners, those who wore traditional shoes had injury rates 3.41 times higher than experienced minimalist shoe runners (46.7% shod vs. 13.7% minimalist) (Goss & Gross, 2012). Additionally, a recent study found after 12 weeks of training in minimalist shoes, intrinsic foot muscles' cross-sectional area and arch stiffness increased (Miller, Whitcome et al., 2014), which may be protective against injuries. These data are encouraging for runners interested in switching.

Other researchers have reported edema or stress fractures occurring in various bones of the foot in runners who switched to barefoot-simulating footwear (Giuliani et al., 2011; Ridge et al., 2013), possibly because plantar pressures increase under the forefoot region

when wearing minimalist shoes (Bergstra et al., 2014; Kernozek et al., 2014). Landing on the forefoot region, as during FFS, loads the metatarsals for a longer period of time relative to RFS. The metatarsals experience a bending moment during walking with compression on the superior surface and tension on the inferior surface (Arndt et al., 2002), which necessitates an opposing moment by the plantar fascia and/or toe flexors to prevent excessive bone stress/strain. Presumably, the same is true in running. So if a hRF runner transitions to a FFS too hastily, increased bone strains and strain rates may occur due to muscle fatigue (Donahue and Sharkey, 1999; Yoshikawa et al., 1994; Arndt et al., 2002; Milgrom et al., 2007), primarily of the small toe flexors. Or the rate of bone destruction may exceed remodeling, evidenced by expansive edema (Ridge et al., 2013), either of which may be the mechanism behind the metatarsal stress fractures Giuliani et al. (2011) and Ridge et al., (2013) observed. The complete transition to new footwear occurred over 10 weeks in Ridge et al. (2013), which may be too short, so perhaps runners whose injuries receded with BF running or minimalist shoes transitioned over a much longer time period or also decreased mileage or intensity.

#### Become more economical

Typically it is also assumed that FFS or BF is more economical, therefore enhancing performance. However, some studies support this (Squadrone & Gallozzi, 2009; Fuller et al., 2014; Burkett et al. 1985; Divert et al., 2008; Hanson et al., 2011) while others do not (Gruber, Umberger et al., 2013; Ogueta-Alday et al., 2014; Warne & Warrington, 2013; Franz et al., 2012; Perl et al., 2012).

#### RFS vs. FFS

Regarding shod RFS versus shod FFS, two studies found that shod hRF were either the same or *more* economical as habitual midfoot/forefoot strikers (hFF) at slower, moderate, and higher speeds (3.06-4.17m/s (Gruber, Umberger et al 2013; Ogueta-Alday et al., 2014). Perl et al., (2012) too found no difference in  $VO_2$  between foot strike styles in experienced minimalist/BF runners. Based on kinetic data, Stearne et al (2014) supported these results and found total average negative and positive power was greater with a FFS. Contrarily, Williams et al. (2012) found the opposite to be true; total power decreased when hRF used a FF. Interestingly there were trends towards total power for hRF runners using a FFS being



greater than hFF runners using a FFS (Stearne et al 2014), so it may be the novelty of the task making hRF runners less efficient.

All these studies have only looked at acute changes, though. Warne & Warrington (2013) came to similar conclusions when trained runners *initially* wore VibramFiveFingers, but after 4 weeks of habituation, running economy significantly decreased (coincident with an increase in use of a FFS), so it may take at least a few weeks for possible economy benefits to present themselves.

The initial theory for FFS being more economical may have come from the thought that greater passive strain energy from the Achilles tendon and ligaments supporting the arch would be utilized with a FFS (Ker et al., 1987; Perl et al., 2012). However, the additional active eccentric muscular contractions of the plantarflexors during the initial half of stance may mitigate passive energy advantages. Since the prevalence of a RFS increases towards the end of a longer race when fatigue may be setting in (Larson et al., 2011), this may also suggest that RFS is a less fatiguing foot strike style than FFS, or that runners are switching foot strikes to use non-fatigued muscles.

#### *Shod vs. BF*

A recent systematic review on running economy and footwear found minor improvements in running economy in light shoes or BF compared to heavier, conventional shoes (Fuller et al., 2014). A viable explanation for the improved running economy when BF is because the weight of a shoe increases the moment of inertia of the foot, and for every 100 g increase in shoe mass, cost of transportation increases ~1% (Frederick, 1984). For example, runners seem more economical in lightweight shoes (~150 g) or BF than typical running shoes (~350 g) ( $\Delta\text{VO}_2 = 1.3\text{-}1.5 \text{ mL/kg/min}$ ) (Squadrone & Gallozzi, 2009; Franz et al., 2012; Divert et al., 2008; Perl et al., 2012). Hanson et al (2011) also found runners were 3.8% more economical when BF compared to shod, but it is possible the BF condition was performed at a slower speed, invalidating the results (Franz et al., 2012). The study by Franz et al (2012) was unique in that they attached the same masses to the foot (150 g, 350 g) as the shoes tested and found  $\text{VO}_2$  was 3-4% *lower* when shod than BF at comparable masses. And since there was no difference between actual BF and 150 g shoes, they concluded that BF running does not decrease oxygen consumption compared to lightweight shoes. It is

important to note measurement error on most metabolic carts can exceed the measured differences, so results should be interpreted with caution.

Additional support includes almost 70% of runners who began to incorporate BF running into their training reported their race times improved (Hryvniak et al., 2014). We cannot ascertain if they ran the races BF, but perhaps the improved race times stemmed from better conditioning of the plantarflexors or increased intrinsic foot muscular with BF training (Miller, Whitcome et al., 2014). Runners who wore minimalist shoes also reported logging more miles and at a faster pace than runners in traditional shoes or BF (Goss & Gross, 2012).

### **Switching footwear typically leads to switching foot strike**

Runners who use a RFS pattern in traditional running shoes typically switch to a MFS or FFS when running in a minimalist shoe or BF because of the discomfort of landing on their heels (Hamill et al., 2011; De Wit et al., 2000; Lieberman et al., 2010). In fact, in one study, all 10 hRF switched to a MFS when BF (Hamill et al., 2011). However, that is not always the case. Williams et al (2012) observed that when 20 hRF ran BF, 8 still used a RFS (40%), 9 a MFS (45%), and 3 a FFS (15%). Similarly, Cheung & Rainbow (2014) observed 20/30 shod runners used a MFS or FFS strike when BF while the other third used mixed foot strike styles. Retaining a heel strike while barefoot may possibly be harmful as several studies noted very high GRFs and LR<sub>s</sub> under such circumstances as compared to shod RFS or BF with a FFS (Cheung & Rainbow, 2014; DeWit et al, 2000; Stacoff et al., 2000; Lieberman et al. 2010; Goss et al., 2015). If switching to minimalist footwear or BF, it may be best to also switch to a MFS/FFS and do so extremely slowly to allow active and passive tissues to adapt to the new loads. A 10-week transition to barefoot-simulating shoes may be too quick (Ridge et al. 2013).

Briefly, it is not just the presence or absence of a shoe that may influence chosen foot strike style but also the surface characteristics. Gruber, Silvernail et al. (2013) found that if hRF ran BF on a hard surface, 65% ran with a MFF/FFS but only 20% ran with a MFS/FFS on the soft surface.

## **Biomechanical differences between foot strike & footwear**

There are some obvious biomechanical differences between foot strike styles or footwear (such as a more plantarflexed foot at contact with a FFS) and other less obvious ones that may explain the differences in oxygen consumption and injury prevalence just discussed. They primarily include changes in spatiotemporal, kinematic, and kinetic parameters and the underlying muscle activity controlling those movements.

### Spatiotemporal

#### *RFS vs. FFS*

Spatiotemporal parameters may be slightly different between shod FFS and RFS. Gruber, Umberger et al. (2013) and Gruber (2012) noted a small effect of SL being slightly shorter (~2-5 cm, or ~1.6-1.8% shorter than PSL) during shod FFS than RFS. The distance between the body's center of mass and the heel was 5.7 cm shorter and there was a small effect size for a shorter step length and medium effect size for increased cadence for hFF (Kulmala et al., 2013); however, there was no difference in step width. Kernozek et al. (2014) did not notice any difference in contact time between hRF and hFF after training in minimalist shoes for 4 weeks without a notable change in foot strike style. After 6 weeks of training, Diebal et al (2012) noted significantly shorter contact time, stride length, and increased cadence in hRF who switched from shod RFS to FFS.

#### *Shod vs. BF*

While the differences are not as large when shod, contact time and step/stride length are significantly shorter during BF running compared to shod (De Wit et al., 2000; Bonacci et al., 2013; Bonacci et al., 2014; Squadrone & Gallozzi, 2009; McCallion et al., 2014; Paquette et al 2013; Kerrigan et al 2009; Shih et al., 2013). Runners also seem to have a slightly longer stride time/longer contact time in traditional shoes versus minimalist shoes (Bergstra et al., 2014; McCallion et al 2014). If running velocity is not held constant, novice barefoot runners tend to self-select a slightly slower running velocity and use a shorter SL as compared to shod running (Thompson et al., 2014). Related to SL, the foot is placed further anterior to the hip at contact when shod (De Wit et al., 2000).

### Kinematic changes

#### *RFS vs. FFS*

By definition, the foot has to be more plantarflexed at contact with a MFS or FFS compared to RFS. As described earlier, with a shod RFS, the foot lands at  $\geq 8^\circ$  relative to the

ground and approximately  $-1.6$ - $8^{\circ}$  plantarflexion with a shod MFS or  $< -1.6^{\circ}$  with a FFS (Altman & Davis, 2012a).

Several authors have noted that shod RF results in a straighter knee at contact (Arendse et al., 2004; Gruber, 2012; Laughton et al., 2003; Lieberman et al., 2010; Shih et al., 2013) and a more flexed hip compared to FFS (shod, minimalist, or BF) (Shih et al., 2013; Willy & Davis, 2014). Williams et al., 2012 did not observe this, but was underpowered to detect differences in hip angle at contact, as it was nearly  $6^{\circ}$  more flexed ( $ES \sim 1.0$ ) for shod RF than BF. The greater knee and hip flexion are proposed to be a mechanism allowing the foot to be more plantarflexed at contact (Shih et al., 2013; Williams et al., 2000). Because the knee is straighter at contact with a RFS, runners tend to go into greater peak knee flexion than FFS (Kulmala et al., 2013; Bonacci et al., 2014; Paquette et al., 2013). Laughton et al. (2003) found that tibial accelerations were significantly higher for FFS than RFS. While this may have implications for injuries, it is more likely a product of greater knee flexion at contact, which decreases the effective mass that is being accelerated (Derrick, 2004).

In the frontal plane, hFF did not have as large of peak hip adduction angle as hRF (Kulmala et al., 2013), which may be important because it is associated with PFPS and ITBS (Ferber et al., 2010; Noehren & Davis, 2007; Wilson & Davis, 2008; Noehren et al., 2007).

#### Shod vs. BF

Foot angle at contact appears to depend in one's habitual foot strike pattern. De Wit et al. (2000) presumably had hRF run shod and BF and noted a less dorsiflexed foot when BF ( $6.4^{\circ}$  versus  $18.0^{\circ}$ ). In runners who presumably were hFF, average ankle dorsiflexion at contact was  $\sim 4^{\circ}$  when shod and  $0^{\circ}$  when BF (Bonacci et al., 2014; Squadrone & Gallozzi, 2009).

Besides the foot being less dorsiflexed at contact for BF, there are other subtle kinematic differences at the foot and lower leg. Eslami et al. (2007) had people run in running sandals and BF and found no difference in tibial rotation, rearfoot eversion excursion, or eversion to tibial internal rotation (although it tended to be lower in BF ( $ES=0.36$ )). Stacoff et al. (2000) also confirmed similar eversion excursion between conditions when using intracortical bone pins. De Wit et al. (2000) noted a less inverted foot at contact ( $\sim 3^{\circ}$ ) when BF, possibly leading to the smaller eversion excursion when BF v. shod that Paquette et al. (2013) observed. These variables are of interest because excessive

amount of motion has been related to shin splints and pain of the patellofemoral joint and Achilles tendon (Clement et al., 1981; Smart et al., 1980; Tiberio, 1987; Vtasalo and Kvist, 1983), and lower eversion to tibial internal rotation (i.e.g, more tibial rotation relative to the rearfoot) was more prominent in those with knee-related injuries (McClay and Manal, 1997; Williams et al., 2001). Peak eversion velocity, which may also be associated with injury (Messier and Pittala, 1988), was smaller in BF than shod running (Stacoff et al., 2000).

Moving up the chain, the knee is more flexed at contact for BF v. shod (De Wit et al., 2000; Schutte et al 2013; Leiberman et al., 2010; Olin & Gutierrez, 2013; Perl et al., 2012). It is also more flexed in minimalist shoes compared to traditional shoes (Willy & Davis, 2014). Related to this, the shank is more vertical at impact when BF ( $96.8^\circ$  vs.  $99.8^\circ$ ) (De Wit et al., 2000). Interestingly, the opposite is true at the time of first vertical GRF peak, when the shank is closer to vertical for shod. Following that, at midstance, there is approximately  $2\text{-}3^\circ$  *less* peak knee flexion for BF v. shod, and it peaks earlier in stance (De Wit et al 2000; Bonacci et al 2013; Paquette et al 2013; Olin & Gutierrez, 2013; Perl et al., 2012). Peak knee flexion is of interest because it highly correlates with knee extensor moment and patellofemoral contact force (PFCF) during shod running (Lenhart et al., 2013), and for every  $5^\circ$  increase in peak knee flexion, oxygen consumption may increase 25% (Valiant, 1990). This may also partially explain why some researchers found BF running is more economical than shod.

Bonacci et al (2013 shoes) did not observe any kinematic differences at the hip between shod and barefoot. Kerrigan et al. (2009), did, however—hip abduction and internal rotation angles were larger when shod.

BF may result in greater overall lower extremity stiffness compared to BF (De Wit et al 2000) because of smaller joint ranges of motion, although others have found no difference (Shih et al., 2013).

### Ground reaction forces

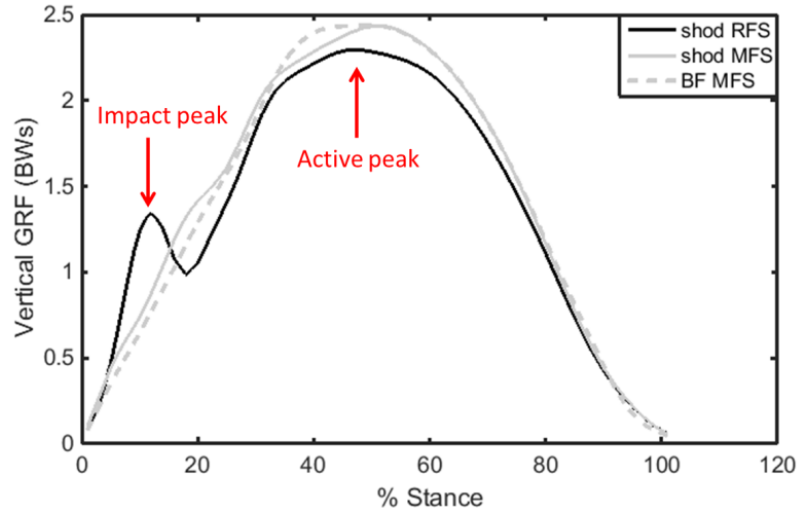
#### *RFS vs. FFS*

The primary kinetic variable most researchers focus on when comparing RFS and FFS is the vertical ground reaction force (GRF) because of the distinct difference between the two. Shod FFS is characterized by an absence or minimization of a vertical impact peak (VIP) in the vertical GRF compared to RFS (Fig. 2) (Cavanagh and LaFortune, 1980; Nilsson

and Thorstensson, 1989; Oakley & Pratt, 1988; Giandolini et al., 2012). However, MFS may still show an impact peak (Hamill et al., 2011; Gruber, 2012; Squadrone & Gallozzi, 2009). The impact peak occurs at a frequency of ~10-20 Hz (Derrick et al., 1998; Hamill et al., 1995; Nigg, 2001) and the active peak, or second peak, is associated with frequencies below 8 Hz (Potthast et al., 2010; Shorten & Mientjes, 2011) (See Figure 2). A few authors have noted that the active peak is larger for FFS than RFS (Oakley & Pratt, 1988; Altman & Davis 2012b; Gruber, 2012; Kulmala et al., 2013), although it appears to be lower in BF than shod (Squadrone & Gallozzi, 2009; Paquette et al., 2013; Thompson et al., 2014; Divert et al., 2005; De Wit et al., 2000; Kerrigan et al., 2009). Peak force measured using plantar pressure insoles, which is slightly different than GRF, was not different between hRF and hFF runners when wearing minimalist shoes (Kernozek et al., 2014).

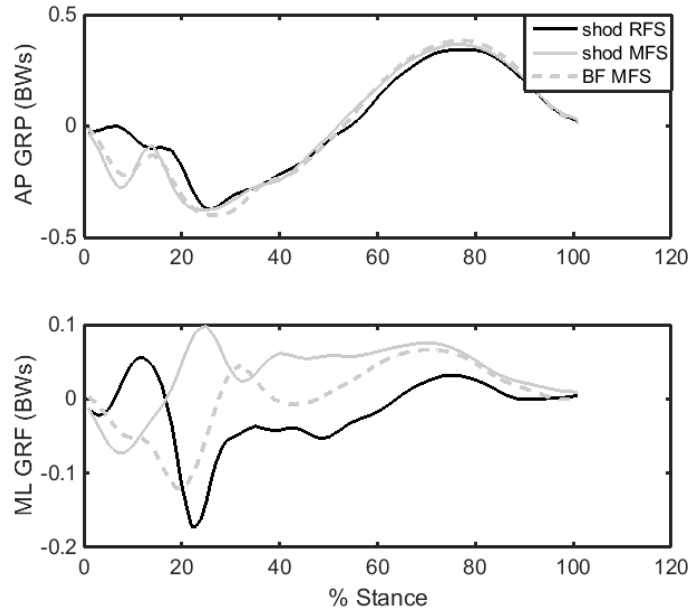
Vertical loading rate (VLR), or how quickly that vertical GRF increases, may be lower for FFS versus RFS (Cheung and Davis, 2011; Diebal et al., 2012; Giandolini et al., 2012; Shih et al., 2013) or similar between foot strike styles (Laughton et al. 2003), although these five studies all used converted (i.e., told hRF runners to run with a FFS) rather than habitually shod mid/forefoot strikers. Only a few studies have used habitual foot strike groups, but they too found lower VLR in hFF compared to hRF (Gruber, 2012; Kulmala et al., 2013; Oakley & Pratt, 1988). It is important to note in Gruber (2012) that midfoot strikers who exhibited an impact peak in the vertical GRF profile were classified as hRF, while those without an impact peak were considered hFF. Several retrospective (Hreljac et al., 2000; Milner et al., 2006; Zifchock et al., 2006; Pohl et al, 2008; Pohl et al., 2009) and prospective studies (Bowser et al., 2010) and a recent review (Zadpoor and Nikooyan, 2011) found that increased VLR is associated with stress fractures, which is likely why FFS may decrease injury risk. However, to my knowledge, all the studies that have found elevated VLRs in injured runners were hRF (Hreljac et al., 2000; Milner et al., 2006; Zifchock et al., 2006; Pohl et al, 2008; Pohl et al., 2009; Bowser et al., 2010), so it remains to be seen if this association with injury holds true for habitual and converted mid/forefoot strikers.

Despite an apparent elimination of a VIP and lower LR with a FFS, higher impact frequencies are still present in the frequency domain (Gruber, 2012). As demonstrated by Shorten and Mientjes (2011), there may be a vertical impact in FFS but it may not show up as a peak in the time domain.



**Figure 2.** Representative vertical GRF (in multiples of body weight (BW)) for a habitual rearfoot striker. The impact and active peaks are identified. Stride length was the same for all conditions. RFS: rearfoot strike, MFS: midfoot strike, BF: barefoot.

While the magnitude of the vertical GRF during running is much larger than the other ordinal directions, there are differences in the shear forces between foot strike styles. Posterior and medial impact forces occur in FFS and not in RFS (Cavanagh and LaFortune, 1980; Nilsson and Thorstensson, 1989; Laughton et al., 2003) (Fig. 3). Since bone is weaker in shear than compression (Reilly & Burstein, 1975; Turner et al., 2001; Hayes & Bouxsein, 1991), these higher forces in FFS versus RFS should not be ignored. Milner et al. (2006) did not find a retrospective association between posterior LR and stress fracture history in hRF, so it may not be a dominative factor in bone stresses. However, posterior GRF is lower in RFS than FFS, so it the larger value for FFS may play a role in injuries.



**Figure 3.** Representative anterior-posterior (AP) and mediolateral (ML) GRF (in multiples of body weight (BW)) for a habitual rearfoot striker. Stride length was the same for all conditions. Positive values correspond to anterior and lateral. RFS: rearfoot strike, MFS: midfoot strike, BF: barefoot.

#### Shod vs. BF

As mentioned briefly, retaining a RFS while BF may induce very high loading rates, but running barefoot with a FFS resulted in similar VLR relative to shod RFS (Leiberman et al., 2010). VLR while BF is approximately 4-7 times higher if maintaining a RFS compared to shod RFS or barefoot FFS (De Wit et al, 2000; Leiberman et al., 2010). Since runners used a RFS when BF in De Wit et al. (2000), the VIP and peak active vertical GRF were not significantly different between than shod RFS, but usually more than one impact peaks were observed when BF. If hRF do switch to a MFS/FFS when BF, a reduction of about 0.2-0.3 BWs is observed for the VIP, but occurrence of that peak was much sooner than shod (Hamill et al., 2011). Despite earlier occurrence, the average VLR was about half as large for BF FFS than shod RFS. Perhaps different results would have been obtained if peak instantaneous VLR was compared instead.

Peak active vertical GRF was larger or tended to be larger when shod versus BF (Kerrigan et al., 2009; Paquette et al., 2013; Thompson et al., 2014) or in minimalist shoes (Paquette et al., 2013). Peak posterior and anterior GRF were larger when BF versus shod



(Kerrigan et al., 2009; Paquette et al 2013). Thompson et al. (2014) did not note significant differences between anterior-posterior or medial-lateral GRF, although all runners may have maintained a RFS in both conditions (evidenced by the continued presence of an impact peak in the vertical GRF curves) whereas some runners in Kerrigan et al. (2009) and Paquette et al. (2013) may have switched to a MFS/FFS when BF.

Interestingly, Willy & Davis (2014) observed after 10 minutes of treadmill running in minimalist shoes, hRF made slight changes in their sagittal plane kinematics but VIP and VLR still increased. The authors felt that the lack of proprioceptive information in the minimalist shoes as compared to true BF running may account for these potentially injurious landing mechanics. Therefore, a similar study found that instructed feedback on transitioning to a FFS when BF was beneficial in decreasing GRF loading variables (Samaan et al., 2014).

#### Loading at the knee

##### *RFS vs. FFS*

Since knee injuries are most common in runners (Taunton et al., 2002; 2003), additional loading is not favorable. With that logic, RFS may seem worse than FFS since knee extensor moments, energy absorption, and patellofemoral loading is greater in RFS, whether shod or BF (Kulmala et al., 2013; Bonacci et al, 2014; Vannatta & Kernozek, 2015; Williams et al, 2012; Goss et al 2015; Paquette et al., 2013). The lower loads at the knee with a FFS likely were the mechanism behind the reduced pain in hRF with patellofemoral pain who switched to FFS (Cheung and Davis, 2011). Interestingly, when hRF ran with a FFS, they exhibited greater knee loads and total lower limb average power compared to hFF when running at the same speed (Stearne et al., 2014; Williams et al., 2000). Therefore, if hRF switch to a FFS, they may not lower their knee loads to the same extent as hFF (Gruber, Silvernail et al., 2013). Rooney & Derrick (2013) compared joint contact forces, which contains estimates of muscle forces (including co-contraction, which is neglected with net joint moments) and found a trend towards greater values with a *FFS*. This highlights the importance of considering muscle forces also to get a better estimate of joint loads.

Frontal plane peak knee abduction moment, a variable associated with knee osteoarthritis pain development and progression (Amin et al., 2004; Miyazaki et al., 2002) was 24-105% lower in shod hFF versus hRF (Kulmala et al., 2013; Stearne et al., 2014). This further supports lower loading occurring at the knee with a FFS.

### Shod vs. BF

Comparing footwear, knee extensor moment was greatest when wearing traditional shoes compared to both BF and minimalist shoes in both hRF and hFF runners (Paquette et al 2013). Likely due to less peak knee flexion, peak knee extensor moment is approximately 8% lower (ES=0.7), and peak patellofemoral contact force decreased 11.6% while running BF versus shod RFS (Bonacci et al., 2014). Knee extensor moment, negative work or power, and positive power generation were greater when shod versus BF (Bonacci et al., 2013; Kerrigan et al 2009; Paquette et al 2013; Sinclair, 2014; Thompson et al., 2014), and knee abduction moment was larger when shod (Kerrigan et al. 2009; Thompson et al., 2014). Kerrigan et al (2009) noted larger external rotation moments when BF, but Thompson et al. (2014) did not. Conflicting results may stem from Kerrigan et al (2009) not using constant SL whereas Thompson et al. (2014) did. Collectively, these data tend to support a decrease in loading at the knee when BF versus shod.

### Loading at the foot and ankle

#### RFS vs. FFS

While loading at the knee decreases with a FFS, loading of the plantarflexors/Achilles tendon and plantar fascia increases (Almonroeder et al., 2013; Braunstein et al., 2010; Williams et al., 2012; Kulmala et al., 2013; Rooney & Derrick, 2013; Goss et al., 2015; Oakely & Pratt, 1988; Sinclair, 2014). Interestingly, maximum plantar pressure force may not differ between foot strike styles but peak plantar *pressure* under the total foot is larger with a FFS (Kernozeck et al., 2014). Dividing the foot into regions, greater loading occurred under the metatarsal for FFS and in the midfoot and heel for RFS. Not all loading about the ankle is higher with a FFS---a dorsiflexion moment at contact is absent with a FFS because tibialis anterior is not required to eccentrically lower the forefoot after ground contact, as it does with a RFS. Less reliance on tibialis anterior, thus, was a likely explanation for 10 hRF military recruits with anterior compartment syndrome not needing surgery after switching to a FFS (Diebal et al., 2012).

### Shod vs. BF

Ankle plantarflexion moment was greater when wearing minimalist shoes compared to both BF and shod RFS in both hRF and hFF runners (Paquette et al 2013). When BF, power absorption and generation were larger at the ankle compared to shod (minimalist,

racing flats, or traditional running shoes) (Bonacci et al., 2013). Additionally, ankle internal rotational and ankle inversion moments are larger when BF versus shod. (Bonacci et al 2013 shoes; Thompson et al. 2014). Kerrigan et al., (2009) also found the ankle external rotation moment to be larger. Peak pressure time integral was larger in minimalist shoes versus traditional shoes, without a change in foot strike pattern (Bergstra et al., 2014).

#### Loading at the hip

##### *RFS vs. FFS*

Conclusions about which foot strike loads the hip more is ambiguous, as some have found no differences (Stearne et al., 2014; Rooney & Derrick, 2013) or lower values for FFS (Kulmala et al., 2013; Williams et al., 2012). Kulmala et al., 2013 found no difference in peak hip extensor moments and greater peak hip abduction moment in hRF than hFF, of moderate effect size, likely because one goes into greater hip flexion with a RFS. Larger hip abduction moment may be of concern for RFS since Eskofier et al. (2012) found a strong association between it and runners who developed patellofemoral pain syndrome.

Total sagittal plane lower extremity joint work and power may be lower with a FFS (Williams et al., 2012) or the same as RFS (Stearne et al., 2014). These contradictions may be due to Williams et al. (2012) using hRF and asking them to run with a FFS while Stearne et al. (2014) used habitual foot strike groups. Because work done seems to be distributed differently among the lower extremity joints with different foot strike styles, whether converting foot strike styles is appropriate may depend on which joint/segment is currently experiencing excessive loading.

##### *Shod vs. BF*

Bonacci et al (2013) did not observe any kinetic differences at the hip between shod and BF. Kerrigan et al. (2009), did, however—hip internal rotation moment was significantly larger when shod. Thompson et al (2014) did not find differences in peak hip extension or abduction moment, but did note external rotation moment was larger (ES ~0.44) when shod versus BF (SL held constant). Hip power absorption was greater in BF and shod RFS than shod FFS. Taken together, the increase in ankle joint work with a FFS are smaller than the decrease in knee and hip joint work, yielding smaller total lower extremity joint power absorption for either FFS or BF compared to shod RFS (Williams et al, 2012). It seems,

therefore, BF running results in less work done per step, but because of increased steps per mile, cumulative decrease in work and lower extremity is unknown.

### Muscle activity

#### *RFS vs. FFS*

Logically, one would expect that a FFS requires activation of the triceps surae before ground contact to incline the foot accordingly, whereas a RFS would require preactivation of tibialis anterior. Data support this (Ahn et al., 2014; Yong et al., 2014; Olin & Gutierrez 2013; Shih et al 2013). More specifically, Ahn et al. (2014) found that hFF activated the gastrocnemius 11% sooner and it remained activated 10% longer than hRF. To maintain a straighter knee angle while the biarticular gastrocnemii are firing, rectus femoris seems to have greater preactivation in FFS (Yong et al 2014). During stance, it is activated similarly between foot strikes (Shih et al., 2013). Interestingly, soleus showed significantly more activity in early stance for RFS than FFS (Yong et al 2014). However, since the soleus is eccentrically contracting during early stance for FFS and likely concentrically for RFS, muscle activation does not necessarily correlate to greater muscle forces being produced. Hamstring activation is not as lucid—Yong et al. (2014) noted preactivation of the lateral hamstrings was greater for RFS while Shih et al. (2013) noted a trend towards biceps femoris preactivation larger for FFS. During early stance, the hamstrings appear more active for FFS (Shih et al., 2013; Yong et al., 2014). Shih et al (2013) also noted greater tibialis anterior activation during stance for RFS. At push-off, there did not appear to be differences between foot strike styles. It is important to note that all runners were shod habitual RFS in Shih et al (2013), which may slightly influence the findings as hRF cannot perfectly replicate FFS as used by hFF (Stearne et al 2014, Williams et al 2000, Rooney & Derrick, 2013).

#### *Shod vs. BF*

Olin & Gutierrez (2013) observed greater average medial gastrocnemius activity during BF running using either foot strike style compared to shod RFS. Similarly, Divert et al., (2005) noted greater peak gastrocnemius and soleus activity in BF RFS compared to shod RFS. While Shih et al. (2012) did not make direct comparisons between shod and BF, it appeared that there were no large differences in EMG activity whether shod or BF when foot strike was held constant. However, if we compare shod RFS to BF FFS, trends between foot strike listed above were observed.

### **Converted forefoot strikers differ from habitual forefoot strikers**

Thus far, there has been some conflicting evidence as to whether variables are similar or different between foot strike styles and when BF. Part of the discrepancy may be due to some studies using only hRF who ran with both foot strike styles, or shod and BF. Only a few studies have compared select kinematics and kinetics between habitual and newly converted forefoot strikers (i.e., hRF told to switch to a FFS) (Williams et al., 2000; Rooney & Derrick, 2013; Stearne et al, 2014). Williams et al. (2000), with only 9 recreational runners per group, found greater plantarflexor moment, ankle power absorption, and peak vertical GRF and shorter contact time in hFF than converted FFS. There was no difference in average VLR, anterior GRF, rearfoot and knee kinematics, or eccentric knee loads (Williams et al. (2000)) or axial contact forces (Rooney & Derrick, 2013) between habitual and newly converted FFS; however, several comparisons were not statistically significant because of insufficient sample size but displayed medium effect sizes (ES, Cohen's *d*) or larger (Tables 2 & 3). Please note in both instances hFF self-selected a speed faster than hRF which may slightly confound the comparisons. Loading at the knee remains larger when hRF convert to FFS (Stearne et al., 2014; Williams et al 2000), implying greater quadriceps force and patellofemoral loading. This continuation of using the knee extensors more may be a lingering effect of the motor program for RFS, which uses the knee extensors more and plantarflexors less (Paquette et al., 2013; Stearne et al., 2014). These acute changes when switching to a FFS that may lead to pain reduction, particularly the decrease in VLR and knee moments/power.

Greater ankle inversion moment and eversion velocity in hFF are related to increased injury risk (McClay, 2000; Messier and Pittala, 1988), which may be of concern for hFF since their values are larger than converted FFS strikers (Williams et al., 2000), although these injury-related associations were found in hRF. Stearne et al (2014) found that converted hRF also had increased peak ankle external rotation moment than hFF, which may be of concern since combined in-phase torsional and axial loads decreases cycles to failure in bone (George and Vashishth, 2005). Stearne et al. (2014) also noted a tendency for larger values in hRF running with a FFS than hFF running with a FFS for ankle abduction moment, hip extensor moment, hip abductor moment, negative ankle power, and hip positive power. As such, hRF

may not drastically decrease their risk of injury related to these variables if they switch to a FFS.

There was a small to medium effect ( $d=0.2-0.33$ ) for converted FFs landing with 2-3° more plantarflexion than hFF (Gruber, Umberger et al., 2013; Williams et al., 2000), illustrating that hRF exaggerate a FFS when initially attempting. This may also partly explain why their knee was flexed ~6° more at contact than hFF. Taken together, novices can generally replicate a FFS, but there are still some differences compared to hFF, particularly whether the ankle or knee is performing more work. Runners may need several months of training to acquire biomechanics more similar to hFF.

**Table 2.** Effect size (Cohen's  $d$ ) of statistically non-significant differences between hFF and converted FFS (i.e., hRF) based on data from Williams et al., 2000

	VALR	Run velocity	EV vel	Knee Flex @cont	Knee IR vel	Ant GRF	Ank Invers mom	Knee ext mom	Knee power abs	Knee neg work
<b>Effect size</b>	0.50*	0.45*	0.89*	0.78^	0.57*	0.67*	0.72*	0.77^	0.57^	0.93^

\**hFF > converted FFS*

^ *converted FFS > hFF*

**Table 3.** Effect size (Cohen's  $d$ ) of statistically non-significant differences between hFF and converted FFS (i.e., hRF) based on data from Rooney & Derrick 2013

	Run velocity	Ankle contact force	Knee contact force	Hip contact force
<b>Effect size</b>	0.49*	0.47*	0.87*	0.98*

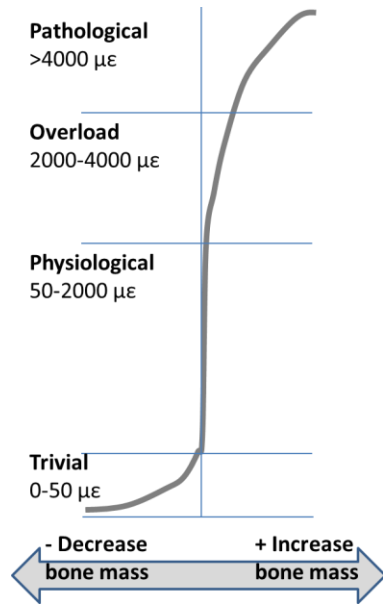
\**hFF > converted FFS*

### Bone health and injuries

The different types of loading mentioned previous may impose beneficial or detrimental loads on the lower extremity bones, primarily depending if adequate rest is allowed. Bones function to attenuate impact energy during running, primarily those at frequencies greater than 18 Hz (Paul et al., 1978). Muscles also help attenuate impacts through eccentric contraction (Derrick et al., 1998; Novacheck, 1998; Winter, 1983). Since the VIP of RFS occurs quickly, it is thought that passive mechanisms are relied on more to attenuate it (Nigg et al., 1981; Williams and Cavanagh, 1987). Deformation of running shoes,

the heel's fat pad, ligaments and articular cartilage are other passive mechanisms for shock attenuation (Chu et al., 1986; Paul et al., 1978). All passive tissues work together to attenuate the high frequency components (Lafortune et al., 1996; Nigg et al., 1981; Paul et al., 1978; Voloshin et al., 1985). As alluded to earlier, hRF had greater power at the higher frequencies (18-43 Hz) while hFF had lower power at these frequencies but higher power at the lower frequencies (1-16 Hz) (Gruber, 2012). Strictly based on the attenuation seen at various tissues, this may imply that hRF may suffer more injuries to passive tissues, like bone, ligaments, and cartilage and hFF may suffer more injuries to active tissue (ie., muscles) if the energy absorbed exceeds tissue tolerance. This hypothesis needs to be substantiated with larger objective studies, as Dauod et al. (2012) had insufficient power to detect differences between groups and Goss & Gross (2012) used self-reported foot strike style.

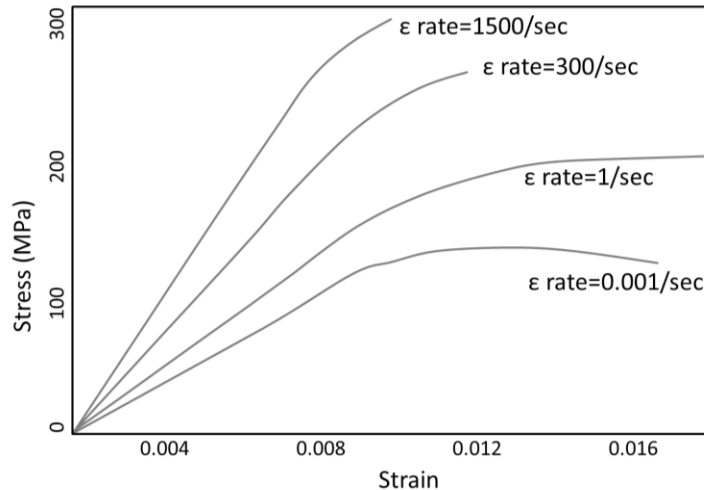
Fortunately, bone is a very living, responding tissue. It can sense the loads placed on it and reorganize its structure (particularly trabecular bone) to better withstand the loads (Lanyon & Rubin, 1984). In adapting to loads, a process called bone remodeling occurs. Osteoclasts are essentially macrophages that resorb old or damaged bone. Osteoblasts are the bone-building cells that lay down new bone. There is a lag of several days between resorption and formation. The entire process of bone remodeling period takes approximately 3-6 months in humans (Bone Remodeling Period, Wikipedia.com). This process is active when a stress fracture is developing, which is a microcrack or fatigue fracture in bone that develops when damage exceeds the rate of bone remodeling (Burr et al., 1990). The lag time associated with this physiological process is part of the reason why stress fractures generally arise within first 2-8 weeks of new physical activity training programs (Brukner, Bradshaw, Khan et al., 1996).



**Figure 4.** Illustration of Frost's Mechanostat Theory indicating strains associated with bone resorption, maintenance, addition, or failure (*adapted from Al Nazer et al., 2012*)

Bone's ability to withstand various loading depends on several things. Externally, the number of loads, their frequency (loading rate), their magnitude, their point of application, their direction relative to the bone, and the phase angle/concurrent loading of different types of loading (George & Vashishth, 2005) all affect fatigability. Frost's mechanostat theory (Frost, 1987) categorizes strain magnitudes as being insufficient, adequate, stimulating, or detrimental to bone health (Fig. 4). It provides a general guideline related to magnitude, but it neglects the other details of the loading environment previously mentioned. Bone fatigue occurs sooner if load magnitude or load rate increases (Carter and Hayes, 1976; Schaffler et al., 1989). Additionally, ultimate stress decreases at lower strain rates, but when strain rates are high, ultimate stress increases (McElhaney, 1966) (see Fig. 5). McElhaney (1996) noted a drastic change at loading rates above 1 Hz.





**Figure 5.** Stress-strain relationship for bone at different loading frequencies (*adapted from McElhaney, 1966*).

Bone is weaker in shear than compression (Beaupied et al., 2007; Turner et al., 2001; Hayes & Buxsein, 1991), so if a constant force is applied at different orientations to a bone's long axis, risk of injury changes. Specifically in the metatarsals, greater shear forces on the metatarsals are not tolerated as well as vertical forces (Arangio et al., 1998). As mentioned earlier, the shear (AP and ML) forces are larger in FFS versus RFS (Cavanagh and LaFortune, 1980; Nilsson and Thorstensson, 1989; Laughton et al., 2003; Kerrigan et al., 2009; Paquette et al., 2013), so the increased force impulse under the metatarsal heads during FFS may have implications for injury. Since more consecutive load cycles causes bone to fatigue quicker and FFS and BF are generally associated with a shorter SL (ie., more steps per mileage) (De Wit et al., 2000; Bonacci et al., 2014; Warne & Warrington, 2013; Bonacci et al., 2013; Squadrone & Gallozzi, 2009; McCallion et al., 2014; Kerrigan et al 2009; Shih et al., 2013), those conditions may predispose runners to more injuries. However, the reduced load magnitude may be more dominating than the increased cycles, as Edwards et al., (2009) found a 10% shorter SL decreased tibial stress fracture probability, and Willson et al., (2014) found an ~16% shorter SL decreased patellofemoral stress impulse per mile in both healthy and PFPS runners. However, shod FFS SL is typically <2% shorter than shod RFS PSL (Gruber, Umberger et al., 2013; Gruber, 2012) and SL when BF/in minimalist shoes is usually only 2.5%-6.4% shorter (De Wit et al., 2000; Bonacci et al 2014; Kerrigan et al.,

2009; Warne & Warrington, 2013; Squadrone & Gallozzi, 2009), so cumulative loading for these smaller differences still need to be ascertained.

Fatigue life of bone is also reduced if it is loaded simultaneously in compression and torsion (George and Vashishth, 2005). In fact, fatigue life was seven times shorter when the two loading modes were applied to bovine tibiae in phase versus out of phase. This has enormous application for running data. It may imply that if peak compression and torsional loading occur more in-phase than out-of-phase, stress fractures may be more likely. The phase angle of these loading conditions has not been investigated in RFS versus FFS or shod versus BF. These variations of loading may partially explain the discontinuity between the cutoffs in the mechanostat theory and observed in vivo tibial strains during running typically not inducing strains  $>2000 \mu\epsilon$  but still causing injury.

There has been some debate whether external forces (e.g., GRFs) or internal forces (e.g., muscle forces) are more important for osteogenic stimulation or bone fatigue (Kohrt et al., 2009). From their review, surmountable evidence for one over the other was lacking. It is estimated tibial loads are greatest around midstance when muscle forces peak (Sasimontongkul et al., 2007; Scott and Winter, 1990; Meardon & Derrick, 2014). More support for muscle forces causing greater loading in bone comes from in vivo strains being greater during a jumping task where the subject landed with a FFS compared to RFS, despite vertical and posterior ground reaction forces being larger for a RFS landing (Ekenman et al., 1998). Similarly, Altman & Davis (2012c) noted greater peak strain rates in shod FFS than shod RFS running, even though VLR are typically lower with a FFS. So while many researchers comparing RFS versus FFS and shod versus BF typically focus on the external GRF and the vertical impact peak in the first 20% of stance, that time may not be the true point in stance when the bone is most susceptible to injury. However, it may bear importance if that initial higher LR preceding the active GRF somehow makes the bone less able to withstand the subsequent larger load (see the two peaks in Fig. 2 during RFS).

Internally, bone geometry, such as cross sectional area (CSA), will dictate the magnitude of the stress while the elastic (Young's) modulus, section modulus, and moment of inertia will determine bone's ability to resist these axial and bending loads (Beck et al., 1996). In general, if these variables are larger, loading decreases. For instance, male marine recruits who developed a stress fracture over 12 weeks of basic training were on average

smaller in stature, but also had smaller tibial CSA, width, and section modulus relative to their body size (Beck et al., 1996). But if bones are given ample time to adapt to the loads, they may not injure as readily.

Finally, the converse train of thought should not be discounted; bones need adequate stimulation of certain strain rates and magnitudes to maintain and improve bone integrity (Qin et al., 1998). Rest between loads is not only important to prevent injury, it also helps restore mechosensitivity, or bone's positive osteogenic response to a load (Robling et al. 2001). As strain magnitude increases, the necessary loading frequency to induce osteogenesis decreases, and vice versa (Hsieh & Turner, 2001). Running with different foot strike styles or BF is a suitable example, since RFS typically has higher LRs but few loading cycles (slightly longer SL) whereas the opposite is true for FFS when BF or shod. Static loads are just as detrimental to bone health (13% decrease) as no loads, highlighting the importance of dynamic loads (i.e., cyclical loads at some frequency) (Lanyon & Rubin, 1984). A threshold of greater than 15 Hz has been suggested as necessary to conserve bone density (McLeod and Rubin, 1990). Based on this cut off, RFS may be better for bone health (if not done in excess of bone remodeling rate), as Gruber (2012) found greater power at 18-43 Hz compared to FFS.

If bones are sufficiently loaded with adequate time for rest, they make several adaptations to loads. For example, Troy et al. (2013) prospectively observed that loading the distal radius three days per week for 28 weeks resulted in increased bone size, mass, and moments of inertia. Similarly, bone mineral content and density showed greatest increases where localized strains were the largest after just 14 weeks of repetitive loading (Bhatia et al., 2015). Interestingly, bone strength or fracture resistance can increase substantially (64-94%) with only minute increases (5-7%) in bone mineral content and density (Turner & Robling, 2003), which primarily occurs due to reorienting trabecular bone.

Taken together, even though hRF might display a VIP or higher LRs than hFF, their bones may have adapted to that load, which may result in similar bone stresses and risk of stress fractures between habitual foot strike groups. However, runners' bones may be more susceptible to injury during the initial months of transitioning to new running styles/footwear, when loads would be changing. If transitioning to a different foot strike or to higher mileage is done slowly to allow plenty of time for bone to adapt, perhaps foot strike is

not a dominating predictive factor for overuse injuries. Other causal mechanisms, such as previous injury history (Buist et al., 2010), excessive training/higher running mileage (Brunet et al 1990, Koplan et al 1982; Macera et al 1989; Walter et al 1989), Q angle  $>15^\circ$  (Cowan et al 1996, Rauh et al 2006; Shaffer, 2006), small stature and bone size (Beck et al., 1996), or inadequate diet, etc. should not be discounted.

### **Bone loading profiles during running**

During running, the GRFs, joint moments, and muscular forces predominantly load the tibia in axial compressive and bending (Sasimontongkul et al., 2007; Scott & Winter, 1990; Ekenman et al., 1998). The bending loads superimpose on the compressive loads on the posterior surface of the tibia to further increase compressive stress and decrease the compressive load on the anterior surface so that it predominantly experiences tensile stresses (Ekenman et al., 1998; Meardon & Derrick, 2014). The compressive forces are vastly due to muscle contractions, while shear joint contact forces are thought to chiefly be caused by shear GRFs or orientation of the tibia relative to the ground (Sasimontongkul et al., 2007).

#### Joint contact forces

Joint reaction forces are forces that occur at a joint without consideration of additional muscle forces (i.e., they are a result of gravity and segment accelerations), whereas joint contact forces do consider the contribution of the muscle. Reaction forces are generally smaller than muscular forces (e.g., ankle reaction:  $\sim 2$  BW, muscle force:  $\sim 7.2$  BW during running at 3.5-4 m/s (Sasimontongkul et al., 2007)). Other estimates for peak contact forces range from 9-14 BW for the ankle (Scott & Winter, 1990; Burdett, 1982; Glitsch & Bauman, 1997; Rooney & Derrick, 2013), 12-15 BW for the knee (Rooney & Derrick, 2013; Edwards et al., 2008; Glitsch & Bauman, 1997), and 5-20 BW for the hip, although most were  $<12$  BW (Giarmatzis et al., 2015; van den Bogert et al., 1999; Rooney & Derrick, 2013; Edwards et al., 2008; Glitsch & Bauman, 1997). In vivo values from an instrumented hip revealed peak axial contact forces of approximately 4.2-5.4 BW when running up to 2.2 m/s. Proportionately, estimates of 9 BW are thus not unreasonable for faster running. While muscular contraction increases total forces at the joints, they generally create bending moments in the opposite direction as reaction forces, functioning as a protective mechanism for the bone (Sasimontongkul et al., 2007; Scott & Winter, 1990).

#### In vivo strain gauge data

To avoid the limitations of modeling, in vivo measurements of bone strains have been used. Besides being invasive and possibly getting inaccurate data from poor bondage, strain values can only be obtained from the cortex of easily accessible locations. The periosteum may not actually be where bone strains are the greatest, though, since some stress fractures occur on the subperiosteal surface (Hayes & Bouxsein, 1991). The earliest in vivo studies of human running was Lanyon et al. (1975) who measured FFS running at 2.2 m/s on the treadmill while BF and shod in 3 participants. Peak strain at toe-off was larger when BF than shod for both principal tensile (+) and compressive (-) strains (+847 v. 746  $\mu\epsilon$  and -578 v. -450  $\mu\epsilon$ ).

Ekenman et al. (1998) applied strain gages to both the anteriomedial and posteriomedial surface of the distal tibia of one female. Peak running values were approximately 1500 and -500  $\mu\epsilon$  in tension (T) and compression (C). During a forward 30 cm jump, different foot strike patterns were used for landing. During the FFS landing, posterior strains were 2700-4200  $\mu\epsilon$ , while it was approximately half as large for RFS landing (1200-1900  $\mu\epsilon$ ). Strains were similar in the anterior and posterior locations during RFS landing but larger on the posterior relative to anterior surface for FFS. These in vivo results corroborate the increased plantarflexor moments accompanying a FFS landing. Several other smaller studies have quantified in vivo tibial strains during running with slightly different protocols. Very few have reported tensile or compressive strains exceeding 2000  $\mu\epsilon$  even at velocities up to 4.72 m/s (Milgrom et al., 2003; Milgrom, Finestone, Levi et al., 2000; Milgrom, Finestone, Simkin et al., 2000; Milgrom, Simkin et al., 2000; Burr et al., 1996). If Frost's mechanostat theory is true, peak running loads are not in the overload or pathological regions, and stress fractures should not occur. However, that is not true, so either the theory or the in vivo data are misleading. I argue that it may both.

#### Musculoskeletal & finite element modeling

The main limitation of strain gauges is that they are typically placed on regions of bone that are easily accessible (e.g., anteromedial tibial surface), thus giving values that are highly location-specific. Using musculoskeletal modeling and finite element analyses, Derrick et al. (2015, unpublished data) has shown that peak compressive stresses in the distal tibia occur in the posteromedial quadrant and peak tensile stresses occur in the anterolateral quadrant (during walking). The same is true for running (Meardon & Derrick, 2014). Both

these regions lie deep to muscles, thus not allowing access for strain gauge adherence. Derrick et al. (2015, unpublished data) estimated stresses at the locations where strain gauges are typically applied and noted they were lower than actual peak values. This is especially true for peak compressive values from gauges placed on the anteromedial tibia since the anterior tibia primarily undergoes tension due to bending about a mediolateral axis induced by plantarflexor contraction. To our knowledge, Ekenman et al. (1998) is the only study to place a gauge on the posteromedial tibia. Although the anteromedial gauge and posteromedial gauge were placed at different levels (mid shaft and distal third, respectively), the peak compressive stresses for the five different activities measured were 16-76 times larger on the posteromedial surface versus anteromedial. But even during the jumping conditions, peak compressive strains did not exceed 2000  $\mu\epsilon$ , which may indicate it was placed closer to the neutral bending axis where strains would be smaller. The posteromedial surface of the distal third of the tibia is where stress fractures typically develop in long-distance runners (Brukner et al., 1998; Crossley et al., 1999; Johansson et al., 1992; Milgrom et al., 1985; Nattiv et al 2013), so it is imperative peak stress values be measured or estimated at this site to better assess stress fracture risk.

#### RFS vs. FFS and Shod vs. BF

Surprisingly, no one has compared individualized bone stresses or strains between habitual foot strike groups when shod or BF, which is what my final study addressed. Preliminary work (Derrick et al., 2012) found tibial stresses were larger in shod hFF versus hRF, but they used a generic bone model (two ellipses). Because of the more triangular-shaped tibia at the distal third (Crossley et al., 1999), an ellipsoid model underestimates peak stresses, especially tensile stresses (Derrick et al., 2015, unpublished data). Other preliminary work did use individualized CT scans to estimate tibial strain and strain rate (Altman & Davis, 2012c), but they only recruited five hRF and had them run at 3.5 m/s with a RFS, FFS, and BF. They found similar peak strains in all three conditions (trend towards larger compressive stresses with a FFS), but strain rate was higher in shod FFS versus shod RFS or BF. Their peak stresses were approximately -10,000  $\mu\epsilon$  and 5,000  $\mu\epsilon$  for compression and tension at the midshaft (Altman & Davis, 2012c). This is higher than in vivo data. Assuming these strains were in the elastic region and the elastic modulus was approximately 20 GPa (Hoffmeister et al., 2000; Rho et al 1993), these strains are equivalent to approximately 200

and 100 MPa, respectively. Considering that typical ultimate compressive and tensile stress for human cortical bone are reported to be ~166 and 108-150 MPa, respectively (Whiting & Zernicke, 2008, 2nd ed), these values may seem high. But since physiological strain rates observed during running are approximately 0.2-0.3 Hz (Altman & Davis, 2012c), ultimate compressive strength increases by 1.33-1.47 times than “static” (0.001 Hz) testing values, or ~200-221 MPa for the femur sample they tested (McElhaney, 1996). Therefore, we believe Altman & Davis’s estimates are reasonable.

The limitations of Derrick et al (2012) and Altman & Davis (2012c) may result in spurious findings, because bone adapts to the dynamic loads placed on it (Lanyon & Rubin, 1984). Specifically, only Altman & Davis (2012c) used subject-specific bone geometry and properties, but they had hRF run with both foot strike styles. Rearfoot strikers’ lower extremity bones may be different than mid/forefoot strikers’ bones because of different GRFs and muscle forces (Cavanagh & LaFortune, 1980; Giandolini et al., 2012; Rooney & Derrick, 2013; Shih et al., 2013). Therefore, stresses and fracture risk may actually be similar between habitual foot strike groups. My final study empirically addressed these limitations by using subject-specific bone models to determine if stress is reduced with one foot strike or the other or when shod versus barefoot, and if stresses change with a shortened SL.

### **Alternative to switching foot strike/footwear: Shorter strides**

Gait retraining may be a more appropriate way to potentially reduce injury risk rather than switching to a foot strike pattern that may place additional stress on maladapted structures and cause injury (Giuliani et al., 2011; Ridge et al., 2013). Several researchers have shown that vertical GRFs and loading rate, AP GRF, tibial accelerations, tibial stress fracture probability, patellofemoral loading, sagittal ankle and knee moments and eccentric work in hRF have been decreased using biofeedback or increasing step frequency/decreasing stride length (Crowell and Davis, 2011; Derrick et al., 1998; Derrick et al., 2000; Edwards et al., 2009; Heiderscheit et al., 2011; Hobara et al., 2012; Lenhart et al., 2013; Thompson et al., 2014; Wellenkotter et al., 2014; Willson et al., 2014). Only one study, to our knowledge, purported that a 10% shortened stride alone during shod RFS did *not* significantly decrease VALR (Cohen’s  $d \approx 0.32$ ), but switching to a midfoot strike while maintaining the same stride length did (Cohen’s  $d \approx 2.0$ ) (Giandolini et al, 2012). The lack of significance for shortened SL are likely overshadowed by the small sample size ( $n=9$ ). Another limitation of

these studies is that they only observed the effects of altering SL in hRF (Derrick et al., 1998; Derrick et al., 2000; Giandolini et al., 2012; Heiderscheit et al., 2011) or they did not specify foot strike style used (Edwards et al., 2009; Lenhart et al., 2013). These studies provide surmountable evidence that shortening runners' stride length could possibly decrease their chance of sustaining an overuse injury.

Fortunately, shortening one's SL a modest amount ( $<10\%$ ) does not make a runner significantly less economical (Hamill et al., 1995; Cavanagh and Williams, 1982), although there is a trend towards increase oxygen consumption. Based on quadratic curve-fitting, most runners' optimal (i.e., lowest oxygen consumption) SL was 1.7-2.9% shorter than their preferred SL (Cavanagh and Williams, 1982; Connick & Li, 2014) or optimal step frequency was 5.6% faster than preferred, with inexperienced runners choosing step frequencies even further from preferred than trained runners (Hamill et al., 1995). Morgan et al. (1994) identified uneconomical runners, defined as preferred SL (PSL) being  $\geq 5\%$  of leg length away from their optimal SL (based on  $\text{VO}_2$ ) or if  $\text{VO}_2$  differed between optimal SL and PSL by  $\geq 0.5$  ml/kg/min. The researchers successfully retrained the uneconomical runners to use a shorter SL (avg: 7.3%) over 3 weeks. This provides evidence that it may be feasible to implement this gait retraining strategy.

Only one study has quantified runners' subjective feelings towards using a 10% shorter stride, which was reportedly uncomfortably and difficult (Eriksson et al 2011). An intervention is irrelevant if it is not implementable. Therefore, our first study assessed the effects of more modest changes in SL and asked about runners' subjective experiences and feasibility of using shortened SLs (5% and 10% shorter than preferred). Besides assessing if significant reductions in lower extremity loading are accomplished with the modest 5% shorter SL, the 5% is slightly closer to the shorter SL used when BF or in minimalist shoes, possibly teasing out the effects of SL.

Briefly, it is useful to report SL as a percentage of leg length (LL) to account for differences in subject height, but LL has been defined differently study to study. Cavanagh and Williams (1982) and Morgan et al. (1994) considered it from the superior border of the greater trochanter to the floor during barefoot standing. Oakley & Pratt (1988) defined it as the ASIS to medial malleolus. With varying definitions, exact comparisons between studies



are complicated. We chose to define LL as the superior border of the greater trochanter to the floor.

### **Stride interval long range correlations**

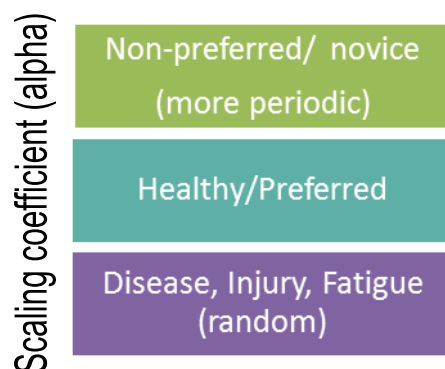
Runners may be at increased risk of injury during their transition to a new foot strike and/or footwear because tissues may be maladapted for the new loads, but also possibly because the novelty of the movement makes the runner more rigid and less capable of adapting (Stergiou & Decker, 2011). More specifically, subtle information from several hundreds or thousands of steps seems to indicate if someone may have a health condition or is moving in a non-preferred manner (Hausdorff et al., 1996). If within a large time-series, such as heart beats or stride times, one value depends on the value of previous points nearby and several points prior, long range correlations are said to exist. The time series demonstrates self-similarity or predictability, as opposed to being white noise. Initially, Peng et al (1995) looked at the heart beat intervals of healthy and unhealthy persons and found a distinct difference. Unhealthy hearts had an overall pattern that was less random and more patterned. An exercise intervention significantly changed the heart beat interval pattern to become more like a healthy heart (Tulppo et al., 2003; Heffernan et al., 2008; Millar et al., 2012). It seemed that some underlying neurological control center/para/sympathetic nervous system may be controlling this that is modifiable with exercise.

Later, Hausdorff et al. (1996) sought to see if this concept could be applied to the human stride interval. Indeed, long range correlations were present in gait. Since then, several researchers have looked at the stride interval patterning in different conditions, like Parkinson's, Huntington's, ALS, elderly fallers, peripheral neuropathy. With PD, Huntington's, ALS, ACL ruptures, and elderly fallers, the patterning was more random than healthy controls (Costa et al., 2003; Hausdorff et al., 1998; Hausdorff, Mitchell et al. 1997; Frenkel-Toledo et al., 2005; Hausdorff, Lertratanakul et al., 2000; Moraiti et al., 2010; Herman et al., 2005). Researchers believe this is unfavorable because the mover is thought to be most adaptable to task or environmental constraints if there is more, but not too much, repeatability (Stergiou & Decker, 2011). Interestingly, peripheral neuropathy (a condition associated with diabetes where sensory nerves in the periphery lose function but supraspinal centers are fully functioning) is not characterized by a change in long range correlations (Gates & Dingwell, 2007). These results exemplify that the underlying control may not be

dependent upon sensory input but is likely controlled in the brain stem, cerebellum, cortex, or deep nuclei. This is not to say that somatosensory information does not affect human movement complexity, since vibrating insoles operating at subsensory levels helped to restore center of pressure complexity in persons suffering from stroke, diabetic neuropathy, and the elderly [Costa et al., 2007; Priplata et al., 2006].

Something that does destroy the natural patterning, though, is the assigned task of walking in sync to a metronome (Costa et al., 2003; Hausdorff et al., 1996; Terrier et al., 2005). Under such constraints, even if the pace is set to preferred stride time, patterning becomes anti-correlated (e.g., a long stride, then shorter stride, longer, shorter, etc.). Other conditions or groups have stride interval patterning that is less random/more predictable, such as speeds above or below preferred, young children, novice runners, and ACL deficient knees (Hausdorff et al., 1996; Hausdorff, Zeman, Peng, Goldberger, 1999; Moraiti et al., 2007; Nakayama et al., 2010). This also seems unfavorable, as being more predictable is thought to be synonymous with less adaptable, so the mover cannot respond to task or environmental constraints (Stergiou & Decker, 2011). In the case of ACL deficient knees, for example, less movement variability may lead to the same regions being mechanically loaded, leading to wear and tear and possibly knee osteoarthritis. If persons do not have the appropriate dynamical degrees of freedom available during these more restricted movements, they may be more likely to get injured. Additionally, there seems to be a systemic affect as the contralateral knee in persons with a ruptured ACL also demonstrated more rigidly compared to healthy controls (Moraiti et al., 2007).

This application has expanded to looking at the stride interval of running (Jordan et al., 2006). Again, at speeds above or below preferred, and in novice runners, stride interval patterning was more predictable than preferred speed and trained runners, respectively (Jordan et al., 2007; Nakayama et al., 2010). Therefore, it seems possible that with supracortical diseases, gait becomes more random; novel or less automated conditions are dealt with by our system in a more periodic sense of control, and; healthy gait lies in the middle (see Figure 6).



**Figure 6.** Spectrum of gait control from random ( $\alpha=0.5$ ) to periodic ( $\alpha>0.5$ ) (*modified from Meardon, 2009*)

More recently, fatigue and a history of previous injury both result in stride interval long range correlations that were more random compared to healthy runners or a less/unfatigued state (Meardon et al., 2011). Perhaps the injury suffered by runners somehow affected their internal clock that controls stride timing, although this directional relationship cannot be determined since it was a cross-sectional study. Cignetti et al. (2009) found a similar effect of fatigue during cross-country skiing. Muscular fatigue has been associated with increased magnitude and rate of bone strain (Donahue and Sharkey, 1999; Yoshikawa et al., 1994; Arndt et al., 2002; Milgrom et al., 2007). If the patterning of loading differs between those with past injuries, perhaps that information can be implemented when doing fatigue tests of bones—e.g., maybe moderately random loading patterns do not cause bones to fatigue/damage as easily as they would when loaded periodically at a set frequency, which is how bone fatigue tests are traditionally done.

Collectively, these results are applicable to our comparison of RFS versus FFS and shod versus BF for a couple reasons. First, performing novel conditions (e.g., switching to a FFS or to BF) may be perceived by the body as a non-preferred movement (similar to running at non-preferred speeds, novice runners, and young children) so stride interval patterns may become more predictable. This may render the runner less adaptable during their time of transition and susceptible to injury. Alternatively, hRF may experience fatigue of the plantarflexors with a FFS, possibly resulting in stride time patterning to becoming more random. Regardless, stride interval long range correlations will likely shift away from their normal, preferred gait.

Briefly, use of these measures provides novel information compared to mean, standard deviation (SD), and coefficient of variation (CV), because the exact same data can be shuffled and still give identical means, SD, and CV as the original dataset but have different fractal properties (Hausdorff et al., 1995), or time series can have similar fractal properties but different means and SD (Gates & Dingwell, 2007). Because of the evidence presented that healthy, preferred running exhibits long range correlations that are not too predictable or too random, it may be beneficial to use it when analyzing novel running conditions.

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**CHAPTER 3**  
**REARFOOT AND MIDFOOT OR FOREFOOT IMPACTS IN**  
**HABITUALLY SHOD RUNNERS**

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**Abstract**

Purpose: Shear loading rates have not been investigated in runners with a mid/forefoot strike (FFS) versus rearfoot strike (RFS). The purpose of this study was to compare three-dimensional ground reaction forces (GRF) and loading rates (LR) during impact in habitual RFS (hRF) and habitual FFS (hFF) strikers. Methods: Thirty competitive runners performed 10 overground running trials with both foot strike styles. Peak three-dimensional and resultant GRFs and instantaneous LRs during impact were compared. Results: Vertical LR significantly decreased for hRF using a FFS (RFS:  $148 \pm 36$ , FFS:  $98 \pm 31$  BW/s) but was similar for hFF running with either foot strike (FFS:  $136 \pm 35$ , RFS:  $135 \pm 28$  BW/s). Posterior impact forces were present during FFS but not RFS, and posterior LR was significantly greater for both groups during FFS ( $-58 \pm 17$  versus  $-19 \pm 6$  BW/s). Medial impact forces were also present during FFS but not RFS, and medial LR was significantly larger for both groups during FFS ( $-21 \pm 7$  versus  $-6 \pm 6$  BW/s). Interestingly, hFF had greater impact peaks and LRs in all directions compared to hRF during FFS. This may be explained by hFF using a smaller strike index (hFF:  $62 \pm 9\%$ , hRF:  $67 \pm 9\%$ ;  $P=0.02$ ), which was significantly inversely related to vertical LR and impact peak. Conclusion: Peak resultant and vertical LRs are not ubiquitously lower when using a shod FFS versus RFS despite an absence of resultant and vertical impact peaks. Furthermore, there were impact peaks in the posterior and medial

directions, leading also to greater LR<sub>s</sub> in these directions during FFS. Therefore, transitioning from RFS to FFS in traditional running shoes may not offer long-term protection against impact-related running injuries since hFF running with a FFS demonstrated many GRFs and LR<sub>s</sub> similar to or greater than RFS.

## **Introduction**

The advent of minimalist shoes and barefoot running has recently caused much hype in the running community. In one study, over 75% of runners surveyed said they were interested in running barefoot or in minimalist shoes, primarily to reduce their chance for injury in the future (Rothschild, 2012). Runners who use a rearfoot strike (RFS) style in traditional running shoes typically switch to a mid- or forefoot strike (FFS) when running in a minimalist shoe or barefoot because of the discomfort of landing on their heels (Hamill et al., 2011). Because switching to minimalist shoes or barefoot usually results in a runner switching foot strike styles, it is sometimes ambiguous whether biomechanical differences are due to changes in footwear, foot strike, or a combination of the two. We are interested in the biomechanical differences between foot strike styles. Few studies have compared foot strike styles while maintaining the same shod condition, which more accurately attributes kinematic or kinetic differences to the foot strike (Cavanagh & LaFortune, 1980, Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013, Laughton et al., 2003, Nilsson & Thorstensson, 1989, Shih et al., 2013). These, therefore, are the only types of studies from which we will draw comparisons.

For runners with patellofemoral pain or anterior compartment syndrome, the reduction or lack of an impact peak (transient peak within the first 20% of stance) in the vertical ground reaction force (GRF) and lower vertical GRF loading rates (LR) during mid/forefoot running may have implications for injury and pain reduction when switching from a RFS to a FFS (Cheung & Davis, 2011, Diebal et al., 2012). One retrospective study has also shown fewer injuries among habitually shod cross country runners with a FFS versus RFS (Daoud et al., 2012) while, adversely, a couple studies noted stress reactions or fractures occurring in runners after switching to barefoot-simulating footwear (Giuliani et al., 2011, Ridge et al., 2013). However, to date, there are no prospective studies comparing injury rates between RFS and FFS runners.

Despite an apparent elimination of a vertical impact peak with a FFS, higher impact frequencies are still present in the frequency domain (Gruber, 2012). As demonstrated by Shorten and Mientjes (2011), there may be a vertical impact in FFS but it may not show up as a peak in the time domain. Regardless, several retrospective studies (Hreljac et al., 2000, Milner et al., 2006, Pohl et al., 2008; Pohl et al., 2009, Zifchock et al., 2006), a prospective study (Bowser & Davis, 2010), and a recent review (Zadpoor & Nikooyan, 2011) found that increased vertical LR (VLR) is associated with running injuries, and VLR has been reported to be lower in shod FFS versus RFS running (Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013). Contrarily, one study (Laughton et al., 2003) did not find differences in instantaneous or average VLRs between foot strike styles when holding shoe condition constant. However, all four of these studies (Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013, Laughton et al., 2003) looked at habitual RFS and asked them to convert to a FFS. Therefore, it is necessary to compare habitual mid/forefoot strikers (hFF) to habitual rearfoot strikers (hRF) to ascertain if the typical differences in kinetics remain. Recent dissertation work by Gruber (Gruber, 2012) has compared average VLR in habitually shod groups and found it to be lower in FFS versus RFS for both groups.

Additionally, bone fatigue increases as the load or load rate increases (Carter & Hayes, 1976, Schaffler et al., 1989), and bone is weaker in shear than compression (Reilly & Burstein, 1975, Williams et al., 2000). As such, it is important to compare the resultant GRF and LR and the three ordinal components between foot strike styles, which has only been done for GRFs but with limited discussion of the shear forces (Cavanagh & Lafortune, 1980, Laughton et al., 2003, Nilsson & Thorstensson, 1989). Greater shear forces on the metatarsal heads are not tolerated as well as vertical forces, likely having implications for injury (Arangio et al., 1998). Landing on the forefoot region, as during FFS, loads the metatarsals for a longer period of time than RFS. The metatarsals experience a bending moment during gait with compression on the superior surface and tension on the inferior surface (Arndt et al., 2002), which necessitates an opposing moment by the plantar fascia and/or toe flexors to prevent excessive bone stress/strain. If a hRF transitions to a FFS too hastily, fatigue of the small toe flexor muscles may occur, increasing bone strains (Donahue & Sharkey, 1999) or rate of bone destruction may exceed remodeling, evidenced by expansive edema (Ridge et al., 2013), either of which may be the mechanism behind the metatarsal stress fractures



observed in converted forefoot strikers (i.e., hRF who switch to a FFS) (Giuliani et al., 2011, Ridge et al., 2013).

Finally, only one study has compared select kinematics and kinetics between habitual and newly converted forefoot strikers (Williams et al., 2000), but variables such as the shear LRs were not investigated. Therefore, the purpose of our study was to compare GRFs and LRs during the impact phase in the three planes of motion during shod rearfoot and mid/forefoot running in healthy competitive runners who habitually run using either a mid/forefoot or rearfoot strike. Based on previously reported similarities between habitual and converted forefoot strikers (Williams et al., 2000) and lower VLR in FFS versus RFS regardless of habitual foot strike (Gruber, 2012), it was hypothesized there would be no differences in the vertical GRFs and LRs between habitual foot strike groups, and VLR will be lower in FFS versus RFS.

## **Methods**

### **Participants**

Fifteen hFF and 15 hRF (3 females in each group) were recruited from a population of competitive runners who were currently injury free and did not report any injury in the previous six months. Midfoot and forefoot strikers and foot strikes were grouped together as forefoot strikers (hFF) or forefoot strikes (FFS) because of similar sagittal ankle kinematics and GRFs observed in our runners and to increase statistical power because of the scarcity of forefoot strikers (Cavanagh & Lafortune, 1980). The study was approved by the Iowa State University Institutional Review Board, and written informed consent was obtained prior to participation. Runners did not differ on age, mass, height, self-selected running speed (described later), or weekly mileage (hRF age:  $20.6 \pm 1.6$  yrs; mass:  $65.7 \pm 7.6$  kg; height:  $1.78 \pm 0.07$  m; velocity:  $4.25 \pm 0.26$  m·s<sup>-1</sup>; mileage:  $59.9 \pm 17.8$  mi/wk; hFF age:  $22.0 \pm 2.3$  yrs; mass:  $63.7 \pm 8.7$  kg; height:  $1.78 \pm 0.08$  m; velocity:  $4.37 \pm 0.23$  m·s<sup>-1</sup>; mileage:  $63.7 \pm 24.6$  mi/wk; all  $P > 0.05$ ).

### **Protocol**

Participants were provided with the same brand and model of running shoes. Anthropometrics were measured and recorded for each participant, including body mass, height, thigh length and circumference, calf length and circumference, foot length and breadth, and malleoli height and width. The self-selected running speed for the treadmill

warm up and all conditions was one that was comfortable and could be maintained for a long run but quicker than an easy recovery run. The same speed was used for both foot strike styles.

A five minute warm up on a treadmill at the self-selected speed was performed before collecting overground running trials. Sixteen retro-reflective markers were placed on the right leg, pelvis, and trunk: fifth metatarsal head, dorsifoot, medial and lateral malleoli, anterior distal and proximal shank, posterior shank, medial and lateral femoral epicondyle, anterior and lateral thigh, both greater trochanters, sacrum, and both acromion processes. Five additional markers were located on the heel counter of the shoe. Marker position data were recorded using an 8-camera motion capture system (Vicon MX, Vicon, Centennial, CO, USA), sampling at 200 Hz. Participants ran down a 30 m runway and were instructed to land with their right foot on either of two adjacent force platforms (AMTI, Watertown, MA), sampling at 1600 Hz. After they performed practice running trials to avoid targeting and allow for natural gait, 10 good trials were collected (no visual targeting, speed  $\pm 5\%$ ).

For the first condition, participants ran naturally with no mention of foot strike style, during which habitual strike style was identified. Strike index (SI) was calculated as the average center of pressure (COP) location during the first 2.5 ms of stance, and reported as a percentage of foot length from the posterior calcaneus. Participants were categorized as hRF (SI < 33.3%) or hFF (SI > 33.3%). Instructions were given on how to run with the opposite strike style and practice was allowed until they felt comfortable. Again participants performed 10 acceptable running trials using the converted strike style; SI was analyzed to confirm the correct foot strike was used.

#### Data Processing

Data processing was performed in Matlab (7.9.0, R2011b). Stance phase was defined as vertical GRF greater than 10 N. Raw kinematic and kinetic data were filtered using a fourth-order zero-lag Butterworth filter at 16 Hz and 50 Hz, respectively. Stance phase data were interpolated to 101 points.

#### Variables

Three-dimensional and resultant GRFs and instantaneous LRs were calculated and normalized to body weight (BW). Peaks were identified from the uninterpolated data during the impact phase. The impact phase for vertical and resultant GRFs and LRs was considered 0-20% of stance, while 0-10% of stance was used for the anterior-posterior (AP) and medial-

lateral (ML) directions based the GRF profiles obtained for mid/forefoot striking (Figure 1). Since there were no distinct impact peaks in the vertical or resultant GRFs for FFS, nor any distinct impact peaks in the AP and ML GRFs for RFS, we selected the value corresponding to the mean time when the impact peak occurred in the opposite foot strike style in that habitual foot strike group (hRF-RFS: 14% of stance for vertical and resultant GRFs; hFF-FFS: 7% of stance for AP and 6% of stance for ML). Therefore, while we may refer to an impact peak for both foot strike styles, it is just the corresponding GRF value in the resultant and vertical directions for FFS and corresponding GRF value in the AP and ML directions for RFS.

### Statistical Analysis

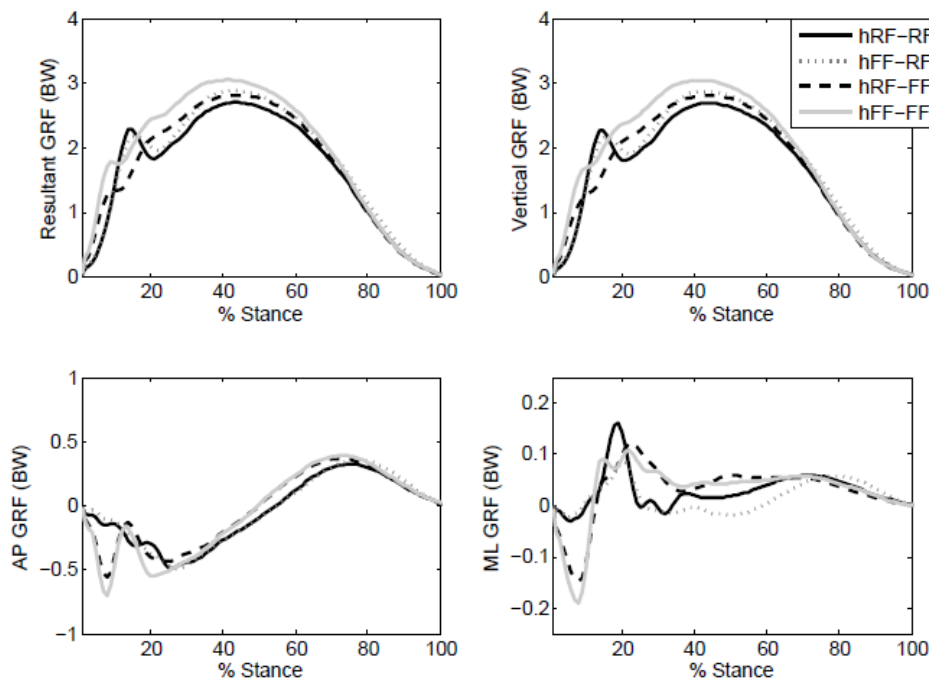
The average of the 10 trials for each condition was calculated and used for comparison. A 2×2 (habitual group × foot strike style) repeated measures MANOVA was run for eight dependent variables: peak impact GRFs and peak instantaneous LR in the AP, ML, and vertical directions and the resultant. Experiment-wise alpha was set to 0.05. If the MANOVA was significant, the univariate ANOVAs were subsequently evaluated at a 0.05 level. Post-hoc Pearson correlation coefficients were calculated to help explain some of the results.

### **Results**

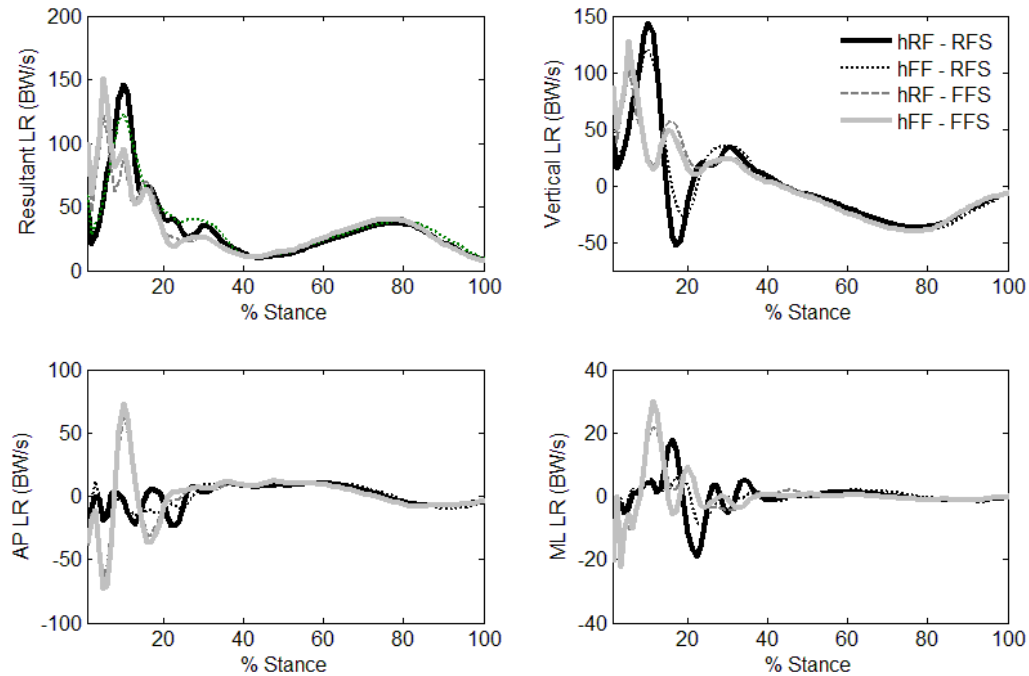
When hRF were asked to run with a FFS, they landed farther forward on their foot than hFF, thus a higher SI (hRF:  $67 \pm 9\%$ ; hFF:  $62 \pm 9\%$ ;  $P = 0.02$ ). Specifically, 11 of the 15 hFF were midfoot strikers ( $SI > 33-66\%$ ) and four were forefoot strikers, whereas seven of the 15 hRF in the converted FFS would be considered midfoot strikers and eight would be forefoot strikers. When hFF ran with a RFS, hFF landed farther back on their heel, resulting in a smaller SI (hRF:  $22 \pm 3\%$ ; hFF:  $20 \pm 5\%$ ;  $P = 0.03$ ). In both instances, the converted group exaggerated the new foot strike style.

Ensemble curves for both the GRFs and LR are shown in Figures 1 and 2. Average peak values during impact are compared in Figures 3 and 4 (supplemental data can be found online). Statistics from the MANOVA revealed a significant habitual group × foot strike interaction ( $P = 0.009$ ), indicating that runners were not able to replicate the impact characteristics of the habitual foot strike style. Univariate tests revealed significant interactions for all eight variables ( $P \leq 0.026$ ). The impact peak of the vertical and resultant

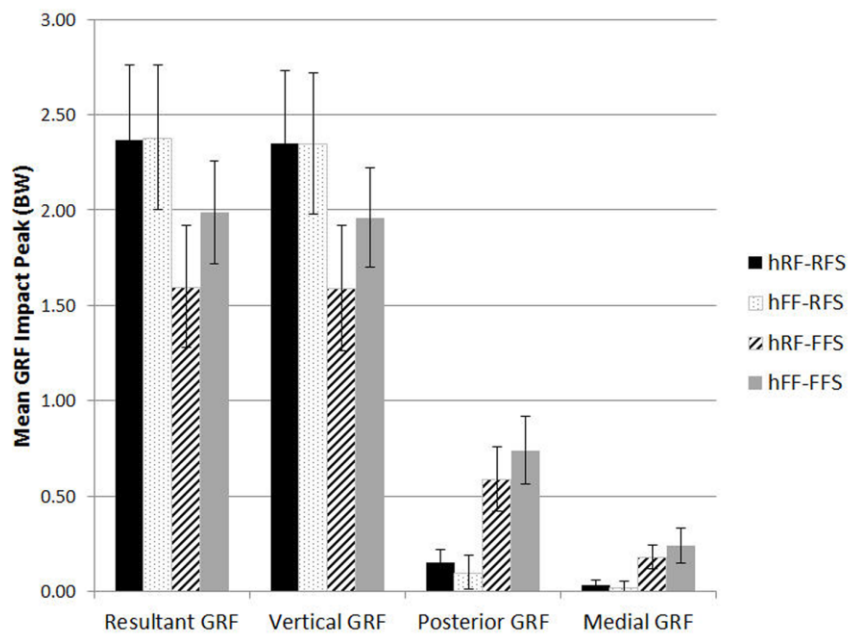
GRFs was similar when both groups ran with a RFS and was decreased/absent for the FFS condition, but they were minimized more when hRF switched from a RFS to a FFS than when the hFF ran RFS versus their usual FFS. When runners used their habitual foot strike, resultant and vertical impact peaks were smaller in hFF versus hRF (Fig. 1). Impact peaks were present in the posterior and medial directions for FFS but not RFS and were approximately 0.5 BW and 0.2 BW higher, respectively, than the corresponding shear GRFs in RFS. These peaks were also larger for the hFF group when running with a FFS versus RFS than when the hRF group ran with a FFS. Resultant LR showed opposite effects depending upon group; it decreased 26% when hRF switched from a RFS to a FFS but was 11% higher when hFF ran with a FFS versus RFS. Similarly, vertical LR decreased 29% when hRF ran with a FFS versus RFS, but it was the same whether hFF ran with a FFS or a RFS. When runners used their habitual foot strike, peak resultant LR was similar and vertical LR was slightly lower in hFF versus hRF (Fig. 2). Posterior and medial LRs increased for both groups when they ran with a FFS, but increased to a greater extent in the hFF group.



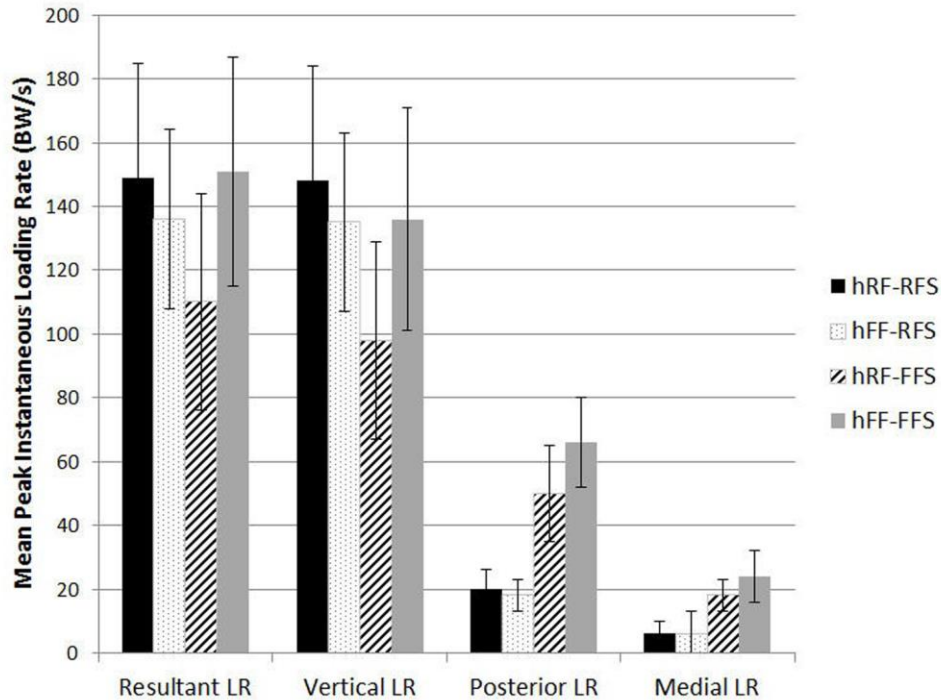
**Figure 1.** Ensemble curves for GRFs for the two habitual foot strike groups (hRF and hFF) during the two running styles (RFS and FFS) normalized to stance time. Positive GRF values represent vertical, anterior, and lateral directions.



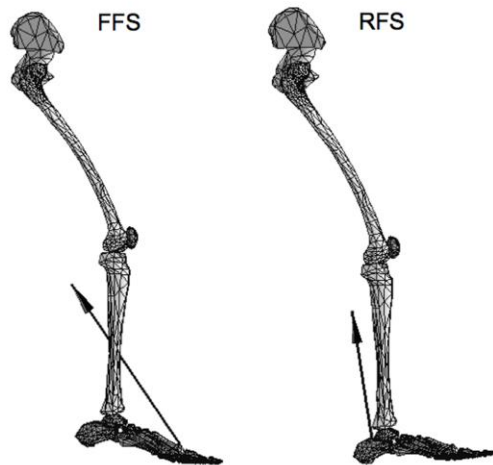
**Figure 2.** Ensemble curves for GRF LRs for the two habitual foot strike groups (hRF and hFF) during the two running styles (RFS and FFS) normalized to stance time. Positive LR values represent vertical, anterior, and lateral directions.



**Figure 3.** Mean GRF impact peaks (with SD bars) for the two habitual foot strike groups (hRF and hFF) during the two running styles (RFS and FFS). All values are represented as positive for comparison. Note: true impact peaks in the resultant and vertical directions were not present for FFS, so the corresponding GRF value at the time when peaks occurred during RFS was used. The same is true for peaks in the posterior and medial directions that were present for FFS but not RFS.



**Figure 4.** Mean peak instantaneous GRF LR<sub>s</sub> (with SD bars) for the two habitual foot strike groups (hRF and hFF) during the two running styles (RFS and FFS). All values are represented as positive for comparison.



**Figure 5.** Representative resultant GRF vector orientation during FFS for a hFF using a FFS (left) and for a hRF using a RFS (right) at 7% of stance (mean occurrence of posterior impact peak for FFS).

The main effect for foot strike ( $P < 0.001$ ) was also significant in the MANOVA. Exploration of the main effect for foot strike is meaningful despite a significant habitual

group  $\times$  foot strike interaction since all changes in variables (besides resultant and vertical LRs) were in the same direction but changed disproportionately between groups. All GRF impact peaks (vertical, posterior, medial, resultant) were significantly different between foot strike style ( $P < 0.001$ ). The resultant and vertical impact peaks were about 32% greater for RFS versus FFS (both  $P < 0.001$ ), but the posterior GRF impact peak was over 450% greater for FFS versus RFS ( $P < 0.001$ ), and the medial GRF impact peak was about 950% greater for FFS versus RFS ( $P < 0.001$ ). Peak instantaneous LR in the posterior direction was 205% greater ( $P < 0.001$ ) and medial LR was 250% greater for FFS versus RFS ( $P < 0.001$ ). There was no significant effect of habitual group ( $P = 0.232$ ).

Post-hoc correlations were run between SI and both vertical GRF impact peak and peak VLR. For both groups running with a FFS, the vertical impact peak significantly decreased as SI increased ( $r = -0.41$ ;  $r^2 = 0.17$ ;  $P = 0.025$ ). SI predicted 18% of the variance in VLR ( $r = -0.423$ ;  $P = 0.020$ ) when both groups ran with a FFS, with decreasing VLR as SI increased.

## Discussion

The purpose of this study was to evaluate the differences between three-dimensional ground reaction forces and loading rates during the impact phase for rearfoot and mid/forefoot running for both habitual rearfoot and mid/forefoot strikers.

### Group $\times$ Foot Strike

Peak impact GRFs (resultant, vertical, posterior, medial) were relatively similar when both groups ran with a RFS but were slightly greater for the hFF running with their natural FFS compared to when the hRF converted to a FFS. Alternatively, the newly converted FFS had lower impacts forces than the hFF. Perhaps this is because when the hRF switched to the novel task of using a FFS, they had to focus more on using a FFS, and in doing so were more aware of the impacts and their neuromuscular system responded by decreasing them. These differences may have also resulted because hRF had a greater SI during FFS than the hFF, and landing further forward on one's foot allowed the plantarflexor muscles to absorb more of the impact energy. We tested this hypothesis by running correlations between SI and vertical impact peak and VLR. The moderate but significant correlations between SI and the two variables indicated that landing further forward one's foot helps lower peak VLR and

vertical GRF impact peak. These data may partially account for the lower vertical impact peak and VLR during FFS for hRF compared to hFF.

Likewise, we found LR<sub>s</sub> (resultant, vertical, posterior, medial) were relatively similar when both groups ran with a RFS but again greater in each direction for hFF versus hRF running with a FFS. Unexpectedly, resultant LR was similar when the two groups of runners ran with their habitual foot strike (~150 BW/s), vertical LR was similar for hFF regardless of foot strike style used (~135 BW/s), and resultant LR was about 40 BW/s greater during FFS versus RFS for the hFF. These results fail to support our hypotheses. The increase in resultant LR is likely a result of greater LR<sub>s</sub> in the posterior and medial directions. Greater VLR during FFS between hFF and hRF groups may partly be due to the smaller SI used by the hFF, which is associated with higher VLR<sub>s</sub>, or perhaps the hRF took precautions to decrease the loading their body experienced during a novel task, as mentioned above. However, in support of our hypothesis and in agreement with previous studies (Cheung & Davis, 2011, Shih et al., 2013), resultant LR decreased 26% and VLR decreased 29% when hRF switched from RFS to FFS, which lies within the approximate 14-32% reduction previously reported for converted forefoot strikers (Cheung & Davis, 2011, Shih et al., 2013).

Taken together, these data suggest that competitive hRF can successfully reduce their vertical and resultant impact GRFs and LR<sub>s</sub> when initially switching to a mid/forefoot strike, which may be beneficial for decreasing risk for running injuries. However, it is possible that after habitually using a FFS, which may lead to a shallower SI, the reduction may be attenuated. This may put habitual mid/forefoot strikers at risk for impact-related injuries. However, to our knowledge, all the studies that have found elevated VLR<sub>s</sub> in injured runners were hRF (Bowser & Davis, 2010, Hreljac et al., 2000, Milner et al., 2006, Pohl et al., 2008; Pohl et al., 2009, Zifchock et al., 2006), so it remains to be seen if this association with injury holds true for converted mid/forefoot strikers. Retrospectively Daoud et al. (2012) found habitually shod competitive RFS had about twice the injury rates of FFS, although this association was minimized and bordered on insignificance (possibly because of lack of statistical power) after removing runners who alternated between foot strike styles. Prospective studies are necessary to substantiate these findings.

For practical purposes, these results indicate that if comparing GRFs or instantaneous LR<sub>s</sub> during RFS, either group of runners may be recruited for RFS, but if the activity of



interest is FFS, it is best to use habitual forefoot strikers. This contradicts Williams et al. (2000) in regards to VLR, although they compared average LR and used recreational runners. Our loading rates are also slightly higher than Williams et al. and others (Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013, Gruber, 2012, Laughton et al., 2003, Milner et al., 2006, Shih et al., 2013, Williams et al., 2000, Zifchock et al., 2006), likely due to the faster running velocities and competitive level of our participants. Similarly, if the intent is to compare SI in FFS, it is best to use the habitual groups.

#### Foot Strike Styles

In agreement with previous literature, there was an obvious vertical impact peak for RFS but not for FFS, and it was larger than the corresponding value for FFS (Cavanagh & Lafortune, 1980, Giandolini et al., 2013; Nilsson & Thorstensson, 1989). However, because of the similar VLRs observed for hFF performing both conditions but decrease for hRF switching to FFS (i.e., a significant interaction), there was not a significant reduction in VLR or resultant LR for FFS versus RFS across groups. This disagrees with previous studies (Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013, Shih et al., 2013), which may likely be because: 1) these studies all used converted rather than habitually shod mid/forefoot strikers, 2) two studies used lower cutoff frequencies of 10 and 20 Hz for kinetic data (Cheung & Davis, 2011, Shih et al., 2013), 3) one study reported average VLR (Giandolini et al., 2013), while Diebal et al. (2012) did not specify if VLR was average or peak instantaneous, 4) or perhaps because all four studies sampled from military recruits or recreational runners and we compared competitive runners. In this case, our data would concur, because hRF had lower VLR when they switched to a FFS from a RFS. Gruber (2012) compared hFF and hRF and found lower rates in the hFF, which conflicts with our findings. This may be because we grouped all midfoot strikers (as defined by SI) into the hFF group, whereas Gruber (2012) grouped midfoot strikers that displayed a vertical impact peak into the hRF group and those without into the hFF group. Regardless, a recent meta-analysis found VLR (but not vertical impact peak) is significantly elevated in runners with a history of stress fractures (Zadpoor & Nikooyan, 2011), possibly making it a more pertinent variable to investigate. To the authors' knowledge, resultant impact peak and LR have not been reported and bears further investigation since the resultant is what the body experiences and is at least as high as or higher than the vertical component.

We also observed that posterior and medial impact forces occurred in FFS and not in RFS, which is consistent with others' findings (Cavanagh & LaFortune, 1980, Laughton et al., 2003; Nilsson & Thorstensson, 1989). The posterior (i.e., braking) impact peak appears to occur in FFS but not RFS because of the rapid change in velocity and direction of the center of mass (COM) of the foot. During a FFS, the COM of the foot is accelerating forward right before foot strike but then slows to a stop and shifts posteriorly as the ankle dorsiflexes after contact before it begins moving anteriorly again during the propulsion phase. This is balanced by an opposing force and shows up as an impact. With a RFS, however, the COM does not reverse directions and travels from posterior to anterior. Similarly, the impact peak in the medial direction in mid/forefoot striking, but not RFS, appears to reflect a faster shift of the center of pressure from the lateral border to midline of the foot after contact (Cavanagh & LaFortune, 1980). This quicker shift may be because our runners landed  $3.5^\circ$  more inverted at contact when mid/forefoot striking than RFS (averaged across group, FFS= $7.7^\circ$ ; RFS= $4.2^\circ$ ).

The posterior and medial GRF component during impact of FFS acts to orientate the resultant GRF vector more parallel to the ground in the first  $\sim 10\%$  of stance (Fig. 5), resulting in greater shear forces, and it is well known that bones are weaker in shear versus compression (Reilly & Burstein, 1975, Turner et al., 2001). Creaby and Dixon (2008) noted that a more medially directed GRF around 30-50% of stance in RFS was found in military recruits who sustained a tibial stress fracture versus controls, speculating that it may have increased the medial bending moment on the tibia because of its typical  $10^\circ$  varus orientation during running (Kawamoto et al., 2002). At midstance, the resultant GRF was slightly lateral for FFS and around zero or slightly medial for RFS, which may be significant given the magnitude of the GRF is greater then, and it is estimated tibial loads are greatest around midstance when muscle forces peak (Sasimontongkul et al., 2007, Scott & Winter, 1990). However, the resultant GRF was significantly more medially directed for FFS versus RFS during the impact phase, which may be more significant given the resultant LR is much greater during this phase and LR seems to have a greater association with tibial stress fractures than GRF magnitude (Zadpoor & Nikooyan, 2011). Additionally, the bones of the foot may also be susceptible to injury as a result of elevated

Because the magnitudes of the shear GRFs and LRs are several times smaller than the vertical magnitudes, their contribution to injury potential may be minute. This notion is supportive of Milner et al. (2006) who did not find a retrospective association between posterior LR and stress fracture history in hRF. However, we found that posterior LR during RFS, as studied by Milner et al. (2006), is over three times smaller than FFS values, so this potential injury association has yet to be established for the much higher shear rates of loading in FFS.

### Implications

Shod rearfoot strikers wishing to decrease their resultant or vertical loading rates may consider switching to a FFS since these variables decreased in our study as well as others (Cheung & Davis, 2011, Diebal et al., 2012, Giandolini et al., 2013, Shih et al., 2013), and elevated vertical LR is associated with a history of stress fracture (Zadpoor & Nikooyan, 2011). These decreases in loading rates, however, may be temporary as hFF running with a FFS had higher loading rates than RFS. Runners must also be aware of the increased shear GRFs and LRs associated with a FFS, which may be important from an injury perspective since bones cannot withstand shear forces as well (Reilly & Burstein, 1975, Turner et al., 2001). If choosing to convert from a RFS to FFS for long distance running (especially if also changing to minimalist footwear or barefoot), it should be done progressively and with caution to avoid injuries (Giuliani et al., 2011, Ridge et al., 2013). As we have discussed, the metatarsals may be more susceptible to injury if converting to a FFS (Giuliani et al., 2011, Ridge et al., 2013) since the mid/forefoot regions are loaded continuously throughout stance and experience greater shear forces at higher loading rates based on our findings. However, shod runners who are plagued by knee injuries or anterior compartment syndrome may benefit by converting (Cheung & Davis, 2011, Diebal et al., 2012). Therefore, whether converting is appropriate may depend on which joint/segment is currently experiencing excessive loading.

Finally, the converse train of thought should not be discounted. Bones need adequate stimulation to maintain and improve their integrity; as strain magnitude increases, the necessary loading frequency to induce osteogenesis decreases, and vice versa (Hsieh & Turner, 2001). So even though each foot strike style and habitual group has its respective larger GRF impact magnitude and loading rate, the runner's bones may have adapted to those

loads and require these forces and loading frequencies to stay healthy. Therefore, while higher loading rates may be associated with bone fatigue and running injuries, other causal mechanisms contribute to an overuse injury (e.g., previous injury history, excessive training, anatomical misalignments, inadequate diet, etc.).

Although the same running shoe was used to abate any effect of shoe type, it is likely that runners with different habitual foot strike styles select different shoes that most appropriately fit their needs. As such, actual GRFs and LR<sub>s</sub> experienced routinely may be different than those represented here, representing a limitation of the study. Secondly, all variables were analyzed in the time domain; slightly different conclusions may have been drawn if analyzing in the frequency domain. Lastly, GRFs and LR<sub>s</sub> only represent the net external loading experienced by the body and not the internal bone loading, which is dominated by muscle forces (Scott & Winter, 1990) and may contribute more to running injuries. Therefore, studies analyzing bone loading environment may be more meaningful when trying to identify injury risk potential between foot strike styles.

## **Conclusion**

In summary, there was an absence of obvious vertical and resultant impact peaks when both groups of competitive runners ran with a mid/forefoot strike versus rearfoot strike but no overall reduction in peak instantaneous resultant or vertical LR<sub>s</sub> across groups. This was because peak resultant and vertical LR<sub>s</sub> during FFS in hFF was still as high as or only slightly lower than the LR<sub>s</sub> of hRF running with their typical RFS. Additionally, impact peaks were present in the posterior and medial GRF profiles when both groups ran with a FFS versus RFS, and the associated shear LR<sub>s</sub> were also greater. When both groups ran with a FFS, hFF had greater GRFs and LR<sub>s</sub> than hRF in all directions, which may partially be explained by a smaller strike index. However, hRF did significantly decrease their resultant and vertical impact peaks and LR<sub>s</sub> when switching to a FFS, but these changes may just be acute effects considering that hFF did not mirror these decreases, assuming both groups of competitive runners are similar. Therefore, switching to a mid/forefoot strike may not result in long-term decreased impact-related injury risk; in fact, given the additional shear impact forces at higher loading rates and reduced corresponding shear strength of bone, injury risk may increase. These findings should be substantiated by prospective studies of impact-related

injuries in mid/forefoot and rearfoot strikers and the internal bone loading environments associated with these foot strike styles.

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## CHAPTER 4

### COMPREHENSIVE JOINT LOADS IN HABITUAL REARFOOT AND MID/FOREFOOT STRIKE RUNNERS WITH NORMAL AND SHORTENED STRIDE LENGTHS

A paper to be submitted to

*Human Movement Science*

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#### **Abstract**

Comprehensive three-dimensional joint contact forces and moments have not been previously reported for shod rearfoot and mid-/forefoot strike running with normal and shorter stride lengths, which is what our study aimed to do. We hypothesized that a rearfoot strike will have higher loads at the knee, mid/forefoot strike will have higher loads at the ankle, while loads at the hip will be similar. Thirty-eight habitual rearfoot and habitual mid/forefoot strikers ran at 3.35 m/s at their normal stride length and 5% and 10% shorter stride lengths. They also ran with the opposite foot strike. Three-dimensional joint moments were calculated at the ankle, knee, and hip. Muscle optimization was used to estimate joint contact forces at the ankle, knee, patellofemoral, and hip joints. In general, loads were higher at the hip joint for rearfoot strike, but higher at the ankle for forefoot strike. The tibia may be under greater loads with a mid/forefoot strike because of the greater axial forces at the ankle and knee (for preferred stride length, *ankle*:  $-10.7 \pm 1.1$  vs.  $-9.3 \pm 1.1$  body weights (BW); *knee*:  $-12.6 \pm 1.2$  vs.  $-11.8 \pm 1.1$  BW). Nearly all variables decreased when using a shorter stride. Four of the five variables that were higher with a rearfoot strike decreased to the same extent whether habitual rearfoot strikers shortening their stride 10% or used a mid/forefoot strike. Different foot strike styles are associated with different loading profiles. Using a certain foot strike style does not collectively decrease all joint loads, so it alone may not decrease risk for all types of running injuries. However, shortening one's stride length unanimously decreased lower extremity contact forces and nearly all joint moments.



## Introduction

There has been a surge of running research on the biomechanical differences between foot strike patterns: rearfoot (or heel) strike (RFS) versus midfoot or forefoot strike (FFS). Most research looking at shod running with different foot strike styles has focused on ground reaction forces. A few have looked at loading at ankle, knee, and hip joints (Kulmala et al., 2013; Rooney & Derrick, 2013; Stearne et al., 2014; Vannatta & Kernozek, 2015; Williams et al., 2012). Nearly all of these have focused on sagittal plane kinetics, while only a couple have reported select frontal and transverse plane variables (Kulmala et al., 2013; Stearne et al., 2014). Only three studies have calculated joint contact forces—Kulmala et al. (2013) estimated compressive patellofemoral (PF) joint forces from net knee extensor moment, Rooney and Derrick (2013) reported axial loads at the ankle, knee, and hip using muscle optimization, and Vannatta and Kernozek (2015) estimated PF joint stresses using muscle optimization. To get a more accurate depiction of joint loads, we must consider muscle forces because they account for the majority of bone loading (Sasimontongkul et al., 2007; Winter & Winter, 1990).

In general, the findings are clear: with a RFS, knee and PF loading is greater than FFS, but with a FFS, there is greater ankle/foot loading (Kulmala et al., 2013; Perl et al., 2012; Rooney & Derrick, 2013; Stearne et al., 2014; Vannatta & Kernozek, 2015; Williams et al., 2012). Conclusions about hip loads are more ambiguous, as some have found no differences between foot strike styles (Stearne et al., 2014), elevated values for FFS (Rooney & Derrick, 2013), or lower values for FFS (Williams et al., 2012). Similarly, total sagittal plane lower extremity joint work and power may be lower with a FFS (Williams et al., 2012) or the same as RFS (Stearne et al., 2014). These contradictions may be due to Williams et al. (2012) using habitual RFS runners (hRF) and asking them to run with a FFS while Stearne et al. (2014) and Rooney and Derrick (2013) used habitual foot strike groups. Habitual FFS runners (hFF) in Rooney and Derrick (2013) also ran slightly faster than hRF, although not significantly, and faster speeds will increase forces. Interestingly, when hRF ran with a FFS, they exhibited greater knee loads and total lower limb average power compared to hFF when running at the same speed (Stearne et al., 2014; Williams et al., 2000). Therefore, if hRF switch to a FFS, they may not lower their knee loads to the same extent as hFF, plus it may not be metabolically advantageous (Gruber et al., 2013).

In regards to the underrepresented frontal and transverse plane data, peak internal knee abduction moment was 24-105% lower in hFF versus hRF (Kulmala et al., 2013; Stearne et al., 2014), and this distinction remained when habitual groups ran with the opposite foot strike (Stearne et al., 2014). Hip abduction moment tended to be slightly higher in hRF (Kulmala et al., 2013; Stearne et al., 2014). Peak ankle internal rotation moment was similar for hRF and hFF, but it increased 33% when hRF used an FFS (Stearne et al., 2014). No other transverse or frontal plane variables have been reported. Since the shear ground reaction forces (GRF) during impact are larger in shod FFS versus RFS (Boyer et al., 2014) and shear forces are not tolerated by bones as well as axial forces (Arangio et al., 1998, Reilly et al., 1975; Turner et al., 2001), it is crucial we further compare loading in these planes.

Finally, several studies, including a systematic review (Schubert et al., 2014), have noted that joint loading or external forces decrease if runners increase step frequency/shorten their stride length (SL) (Derrick et al., 2000; Derrick et al., 1998; Edwards et al., 2009; Heiderscheit et al., 2011; Hobara et al., 2012; Lenhart et al., 2014; Wellenkotter et al., 2014), but adapting such a change may be an imposition for runners. One study found most runners felt a 10% shorter SL was uncomfortable and difficult (Eriksson et al., 2011). As such, we investigated if a 5% shorter SL could significantly decrease loading but be tolerated by runners. Perhaps hRF could achieve similar decreases in loading, particularly at the knee, if using a shorter SL while maintaining a RFS. This may help runners avoid possible injuries associated with transitioning too quickly to a midfoot/forefoot strike and/or minimalist shoes (Giuliani et al., 2011; Ridge et al., 2013). Additionally, only Edwards et al. (2009) and Lenhart et al. (2014) have reported select joint contact forces for shorter SLs, so there is a need to supplement their findings.

Our purpose is to report comprehensive three-dimensional joint moments and contact forces at the ankle, knee, PF, and hip joints during shod RFS and FFS running with normal and shorter stride lengths in both hRF and hFF runners. We hypothesize that contact forces and moments will be greater at the ankle for FFS, greater at the knee and PF joints for RFS, and similar at the hip. With a shorter SL, we hypothesize loading will decrease at all joints.

## **Methods**

### **Participants**

Runners who ran at least 10 miles per week, were injury-free at the time, and did not have lower-extremity surgery were recruited. Forty-two runners completed the running questionnaire and the preferred SL (PSL) conditions for both foot strike styles. However, because of missing markers at the end of stance ( $n=1$ ) and incomplete shorter SL conditions ( $n=3$ ), contact forces were calculated for only 38 runners. They were recreational or competitive runners (19 hRF and 19 hFF) and similar, except hRF had longer legs and were taller, reflecting an unequal number of women (1 in hRF, 7 in hFF) (hRF: age  $21\pm6$  yrs, mass  $72.0\pm10.7$  kg, height  $1.82\pm0.08$  m, leg length  $0.964\pm0.041$  m, weekly mileage  $30.2\pm21.2$  mi; hFF: age  $22\pm3$  yrs, mass  $66.4\pm10.3$  kg, height  $1.74\pm0.09$  m, leg length  $0.921\pm0.058$  m, weekly mileage  $32.9\pm24.0$  mi).

### **Protocol**

The study was approved by the university's Institutional Review Board. Written informed consent was obtained prior to participation. Detailed methods can be found elsewhere (Boyer et al., 2015-in review). Midfoot and forefoot strikers were grouped together as hFF because of the scarcity of forefoot strikers (Cavanagh & LaFortune, 1980). Twenty-five individual reflective markers were placed on the lower extremity and shoes (sacrum, and bilaterally on the ASIS, greater trochanter, lateral thigh, anterior distal thigh, lateral femoral condyle, anterior proximal & distal leg, posterior leg, lateral malleolus, dorsifoot, lateral foot, and heel). Four more markers were placed on the medial malleoli and femoral condyles for the standing calibration trial used to identify knee and ankle joint centers. Participants ran on a treadmill for 5 minutes to warm up before completing seven overground running trials down a 30 m runway, all at 3.35 m/s ( $\pm 3\%$ ). PSL was calculated during the 5 minute treadmill run. All conditions were performed in their own shoes (none were barefoot-simulating footwear). They ran with their habitual foot strike pattern first at their PSL, and then at 5% and 10% shorter SLs. Shortened SL trials were considered acceptable if SL was within  $\pm 2.5\%$  of the target. Overground step length was marked with athletic tape. Strike index (SI) was calculated as the average center of pressure location during the first 2.5 ms of stance after each trial and reported as a percentage of foot length from the posterior calcaneus. A foot strike with an SI  $< 33.3\%$  was a RFS, and FFS had a SI

>33.3%. The only trials included in the analysis were ones with the correct foot strike style based on SI, within the tolerated speed and step length, and no visual targeting. All runners completed a brief questionnaire at the end about their perceptions of the novel foot strike and shortened stride length conditions.

#### Equipment

Kinematic data were collected at 200 Hz with an 8-camera Vicon system (Vicon MX, Vicon, Centennial, CO, USA) and low-pass filtered at 12 Hz. Cardan angles were calculated for the lower extremity segments and joints using a flexion/extension, ab/adduction, internal/external rotation order. Kinetic data were collected at 1600 Hz by an in-ground AMTI force platform (AMTI, Watertown, MA) and low-pass filtered at 20 Hz. SL was calculated as the anterior-posterior distance between the left heel markers.

#### Data Processing and Musculoskeletal Model

Standard inverse dynamics was used to calculate joint reaction forces and moments, which are presented in the distal segment coordinate system. Estimates of segment mass, center of mass location, and moments of inertia were obtained using participants' anthropometrics (Vaughan et al., 1992). Leg length (LL) was defined as the distance from the superior border of the greater trochanter to the floor. SL was expressed as a %LL because of group differences in height. Lower extremity muscle properties were obtained using a musculoskeletal model (Arnold et al., 2010), implemented within custom Matlab software (8.4.0. 150421, R2014b, Natick, MA, USA). Muscle forces were estimated using static optimization and constrained using force-length and force-velocity adjusted maximal values. Because there was insufficient muscle force available to match the peak moments from inverse dynamics, maximum allowable muscle force was increased by a factor of 1.75. A set of muscle forces that matched moments derived from inverse dynamics and that minimized the sum of muscle stresses squared was selected (Glitsch & Baumann, 1997). Sagittal plane moments at the hip, knee, and ankle, and frontal plane moments at the hip and ankle were used for the optimization. Muscle forces were summed with joint reaction forces to estimate joint contact forces.

To estimate PF contact force (PFCF), the patella ligament force to quadriceps force ratio was based on a third-order polynomial curve fit to in vitro data from van Eijden et al. (1987):

$$\text{Ratio} = 2\text{E}^{-07}\text{x}^3 - 0.0001\text{x}^2 + 0.0002\text{x} + 1.15$$

where  $x$  is knee flexion angle ( $^{\circ}$ ). PF contact area was estimated for men and women based on knee flexion angle according to Besier et al. (2005):

$$\text{PF Contact Area (men)} = -0.0242x^2 + 7.3142x + 303.57$$

$$\text{PF Contact Area (women)} = 0.0157x^2 + 4.7478x + 182.95$$

where  $x$  is knee flexion angle ( $^{\circ}$ ) and contact area is given in  $\text{mm}^2$ . PFCF was calculated as the resultant of the quadriceps and patella ligament forces. PF compressive stress (PFS) was found by dividing PFCF by PF contact area.

### Statistical Analysis

Values were averaged across acceptable trials for each condition. A  $2 \times 2 \times 3$  repeated-measures MANOVA (habitual group  $\times$  foot strike  $\times$  SL) was performed in SPSS Statistics 21 to compare three-dimensional joint contact forces at all four joints: ankle, knee, PF, and hip. A second MANOVA compared peak three-dimensional joint moments at the ankle, knee, and hip. Significance was set to  $\alpha=0.05$ . If the MANOVA was significant, the univariate ANOVAs were subsequently evaluated at a 0.05 level. A Sidak adjustment for pairwise comparisons was used. When sphericity was violated, the Greenhouse-Geisser correction was used.

### **Results**

PSL was  $2.50 \pm 0.14$  m for hRF and  $2.36 \pm 0.10$  m for hFF with their normal foot strike styles. Since hRF were taller, PSL as a %LL was no different between groups ( $P=0.598$ ) (hRF:  $259 \pm 12\%$ LL, hFF:  $256 \pm 16\%$ LL). hRF significantly shortened their stride 2.3% when running with a FFS ( $P<0.001$ ) to  $253 \pm 14\%$ LL, but hFF did not significantly lengthen their SL when using a RFS ( $P=0.178$ ;  $258 \pm 17\%$ LL).

The group  $\times$  foot strike  $\times$  SL ( $P=0.108$ ), foot strike  $\times$  SL ( $P=0.169$ ), and group  $\times$  SL ( $P=0.725$ ) interactions were not significant when comparing joint moments, nor was there a group effect ( $P=0.617$ ). There was a significant interaction between group  $\times$  foot strike ( $P=0.037$ ) and significant main effects for both foot strike and SL (both  $P<0.001$ ). For joint contact forces, the group  $\times$  foot strike  $\times$  SL ( $P=0.438$ ), group  $\times$  SL ( $P=0.104$ ), group  $\times$  foot strike ( $P=0.547$ ), and foot strike  $\times$  SL ( $P=0.627$ ) interactions were insignificant, and there was no difference between habitual group ( $P=0.789$ ). There were significant differences for both foot strike and SL (both  $P<0.001$ ).

In general, loading was greater at the hip joint with a RFS, but greater at the ankle for FFS (Table 1, Figures 1 and 2). A shorter SL generally decreased loading at the four joints (Table 2). Comparisons for the habitual groups using both foot strike styles at their PSL are found in Tables 3 and 4.

**Table 1.** Summary of statistically significant results for foot strike style, averaged across group and stride length.

Larger for RFS	Larger for FFS	No difference RFS v. FFS
<b>Knee lateral contact force</b> <b>Hip posterior contact force</b>  <b>Knee extension moment</b> <b>Hip extension moment</b> <b>Hip internal rotation moment</b>	Ankle axial contact force Ankle posterior contact force Knee axial contact force  Ankle plantarflexion moment Ankle internal rotation moment Knee abduction moment Knee internal rotation moment Knee external rotation moment	Ankle medial contact force Knee posterior contact force Patellofemoral contact force Patellofemoral stress Hip axial contact force Hip lateral contact force  Ankle inversion moment Hip abduction moment

**Table 2.** Summary of statistically significant results for stride length, averaged across group and foot strike style (% decrease from PSL for -5%PSL, % decrease from PSL for -10%PSL).

Decreased for -5% or -10%PSL	No difference with SL
<b>Ankle axial contact force (2.0%, 4.5%)</b> <b>Ankle posterior contact force (8.8%, 14.9%)</b> <b>Ankle medial contact force (5.5%, 7.3%)</b> <b>Knee axial contact force (3.2%, 7.0%)</b> <b>Knee posterior contact force (1.2%, 2.8%)</b> <b>Knee lateral contact force (7.3%, 13.9%)</b> <b>Patellofemoral contact force (7.2%, 14.9%)</b> <b>Patellofemoral stress (5.9%, 12.5%)</b> <b>Hip axial contact force (4.5%, 9.2%)</b> <b>Hip posterior contact force (8.1%, 13.4%)</b> <b>Hip lateral contact force (5.0%, 9.7%)</b>	<b>Knee external rotation moment</b> <b>Hip extension moment</b>
<b>Ankle plantarflexion moment (1.8%, 3.8%)</b> <b>Ankle inversion moment (7.2%, 8.4%)</b> <b>Ankle internal rotation moment (7.1%, 9.8%)</b> <b>Knee extension moment (5.7%, 11.8%)</b> <b>Knee abduction moment (2.8%, 5.9%)</b> <b>Knee internal rotation moment (5.1%, 10.7%)</b> <b>Hip internal rotation moment (10.2%, 16.2%)</b> <b>Hip abduction moment (3.4%, 6.4%)</b>	<b>Increased for -5% or -10%PSL</b> <b>Hip flexion moment (-5.4%, -9.2%)</b>

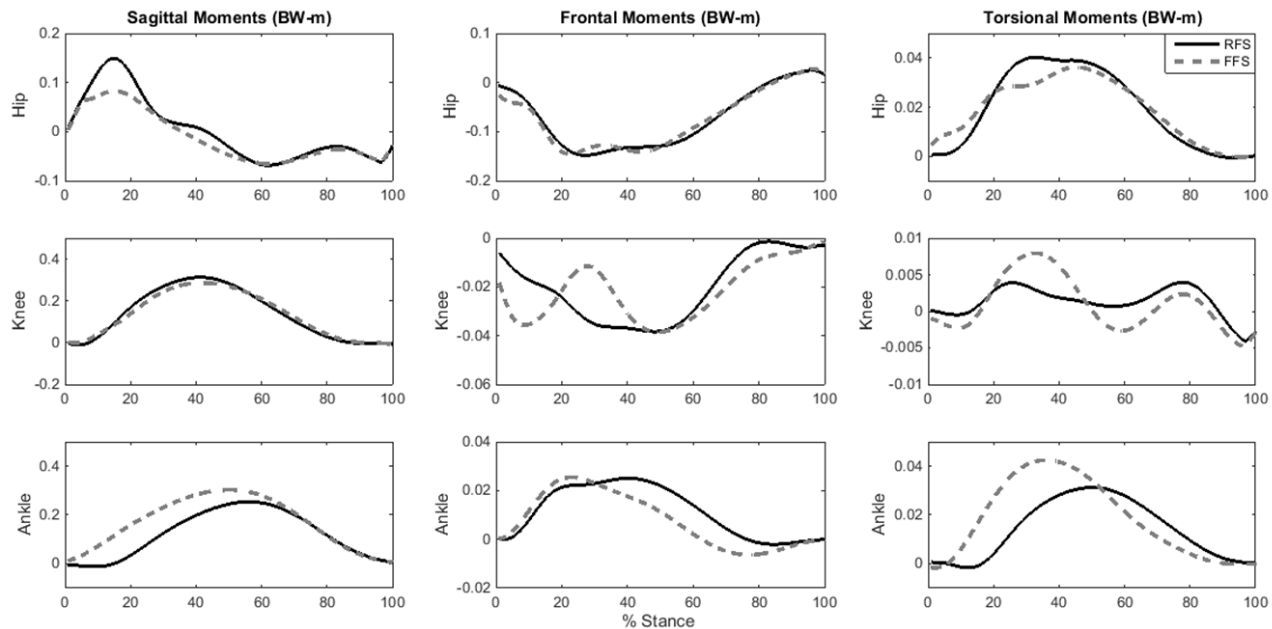
**Table 3.** Peak joint moments (BW·m) during the PSL condition for groups running with habitual and converted foot strike styles (mean  $\pm$  SD).

	ANKLE			KNEE				HIP			
	Plantar-flexion <sup>F</sup>	Inversion	Internal Rotation <sup>F</sup>	Extension <sup>R</sup>	Abduction <sup>F</sup>	Internal Rotation <sup>F</sup>	External Rotation <sup>F</sup>	Extension <sup>R</sup>	Flexion*	Abduction	Internal Rotation <sup>R</sup>
hRF-RFS	0.27 $\pm$ 0.03	0.03 $\pm$ 0.01	0.04 $\pm$ 0.03	0.34 $\pm$ 0.04	-0.055 $\pm$ 0.04	0.010 $\pm$ 0.006	-0.008 $\pm$ 0.007	0.17 $\pm$ 0.03	0.08 $\pm$ 0.02	-0.17 $\pm$ 0.03	0.047 $\pm$ 0.008
hFF-RFS	0.26 $\pm$ 0.04	0.03 $\pm$ 0.01	0.03 $\pm$ 0.02	0.31 $\pm$ 0.04	-0.046 $\pm$ 0.03	0.010 $\pm$ 0.005	-0.008 $\pm$ 0.005	0.16 $\pm$ 0.03	0.10 $\pm$ 0.07	-0.16 $\pm$ 0.03	0.043 $\pm$ 0.016
hRF-FFS	0.31 $\pm$ 0.03	0.03 $\pm$ 0.01	0.05 $\pm$ 0.03	0.31 $\pm$ 0.04	-0.061 $\pm$ 0.04	0.012 $\pm$ 0.009	-0.012 $\pm$ 0.008	0.11 $\pm$ 0.03	0.09 $\pm$ 0.02	-0.18 $\pm$ 0.03	0.045 $\pm$ 0.009
hFF-FFS	0.32 $\pm$ 0.03	0.03 $\pm$ 0.01	0.05 $\pm$ 0.03	0.28 $\pm$ 0.05	-0.051 $\pm$ 0.03	0.013 $\pm$ 0.008	-0.010 $\pm$ 0.007	0.10 $\pm$ 0.03	0.08 $\pm$ 0.05	-0.17 $\pm$ 0.02	0.038 $\pm$ 0.009

<sup>F</sup> Values are significantly larger using FFS

<sup>R</sup> Values are significantly larger using RFS

\*Significant group  $\times$  foot strike interaction



**Figure 1.** Ensemble joint moment curves during stance for PSL averaged across habitual foot strike groups. Positive values represent extension, adduction, and internal rotational moments.

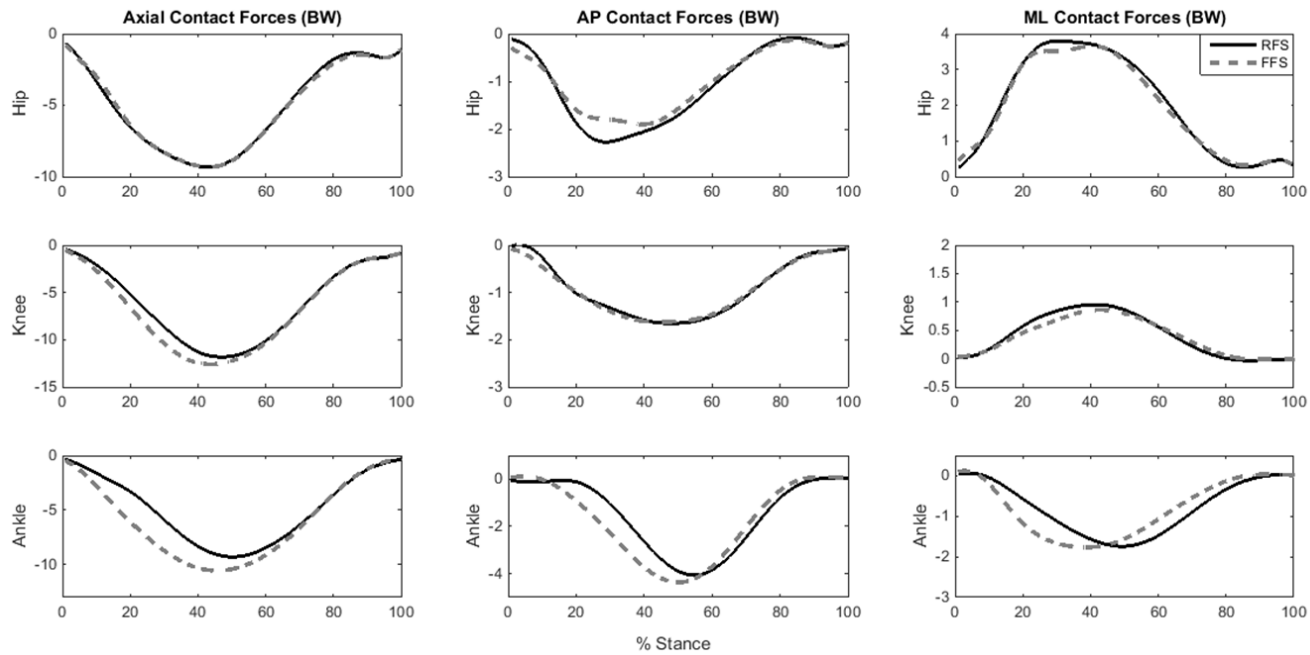


**Table 4.** Peak joint contact forces (BW) and PF stress (MPa) during the PSL condition for groups running with habitual and converted foot strike styles (mean  $\pm$  SD).

	ANKLE			KNEE			PATELLOFEMORAL		HIP		
	<i>Axial</i> <sup>F</sup>	<i>Posterior</i> <sup>F</sup>	<i>Medial</i>	<i>Axial</i> <sup>F</sup>	<i>Posterior</i>	<i>Lateral</i> <sup>R</sup>	<i>Force</i>	<i>Stress</i>	<i>Axial</i>	<i>Posterior</i> <sup>R</sup>	<i>Lateral</i>
hRF-RFS	-9.6 $\pm$ 1.0	-4.4 $\pm$ 1.1	-2.0 $\pm$ 1.5	-12.3 $\pm$ 0.9	-1.7 $\pm$ 0.2	1.0 $\pm$ 0.3	10.8 $\pm$ 1.4	13.9 $\pm$ 2.6	-9.9 $\pm$ 0.9	-2.5 $\pm$ 0.4	4.2 $\pm$ 0.7
hFF-RFS	-9.5 $\pm$ 1.2	-4.2 $\pm$ 1.1	-1.7 $\pm$ 1.2	-12.1 $\pm$ 1.3	-1.7 $\pm$ 0.3	0.9 $\pm$ 0.2	10.1 $\pm$ 1.8	12.1 $\pm$ 2.3	-9.6 $\pm$ 0.9	-2.4 $\pm$ 0.6	4.1 $\pm$ 0.5
hRF-FFS	-10.8 $\pm$ 1.1	-4.4 $\pm$ 1.2	-2.0 $\pm$ 1.3	-13.0 $\pm$ 1.2	-1.7 $\pm$ 0.2	0.9 $\pm$ 0.4	10.8 $\pm$ 1.8	13.8 $\pm$ 3.1	-10.0 $\pm$ 1.0	-2.2 $\pm$ 0.4	4.3 $\pm$ 0.8
hFF-FFS	-11.1 $\pm$ 1.1	-4.8 $\pm$ 1.4	-2.1 $\pm$ 1.4	-13.0 $\pm$ 1.2	-1.7 $\pm$ 0.2	0.8 $\pm$ 0.3	9.9 $\pm$ 2.0	12.0 $\pm$ 2.7	-9.6 $\pm$ 1.0	-2.1 $\pm$ 0.5	4.0 $\pm$ 0.5

<sup>F</sup> Values are significantly larger using FFS

<sup>R</sup> Values are significantly larger using RFS



**Figure 2.** Ensemble joint contact force curves during stance for PSL averaged across habitual foot strike groups. Positive values represent axial up, anterior, and lateral forces. All forces are in the distal segment coordinate system.

### Group $\times$ Foot Strike

Peak hip flexion moment was largest when habitual groups used the opposite foot strike style ( $P=0.002$ ) but were equivalent during their habitual foot strike ( $0.082 \text{ BW}\cdot\text{m}$ ). Although not significant, peak knee external rotation moment was larger when both groups used a FFS, but it increased more for hRF ( $P=0.063$ ), and peak plantarflexion moment was greatest when using a FFS, but it increased more for hFF ( $P=0.077$ ).

### Foot Strike

Peak plantarflexion and ankle internal rotation moments were greater for FFS (both  $P<0.001$ ), while inversion moment was similar ( $P=0.460$ ). Peak axial ( $P<0.001$ ) and posterior ankle contact forces ( $P=0.016$ ) were greater during FFS. Peak medial contact forces were similar between foot strikes ( $P=0.109$ ).

At the knee, peak extension moment was greater for RFS ( $P<0.001$ ), but abduction ( $P=0.001$ ), internal ( $P=0.031$ ), and external rotation ( $P<0.001$ ) moments were greater for FFS. Peak lateral contact force was greater for RFS than FFS ( $P\leq 0.001$ ). Peak axial force was larger for FFS ( $P<0.001$ ). Peak posterior knee contact force ( $P=0.797$ ), PFCF ( $P=0.198$ ), and PFS ( $P=0.197$ ) were similar between foot strike styles.

At the hip, peak extension and internal rotation ( $P\leq 0.001$ ) moments were greater for RFS. Peak hip abduction moment bordered on being significantly greater for FFS ( $P=0.087$ ), and hip flexion was not significant ( $P=0.806$ ) due to the interaction. Peak axial ( $P=0.918$ ) and lateral ( $P=0.222$ ) contact forces were similar between foot strike styles, while peak posterior force was greater for RFS ( $P<0.001$ ).

### Stride Length

At the ankle, peak plantarflexion ( $P<0.001$ ), inversion ( $P=0.003$ ), and internal rotation ( $P<0.001$ ) moments decreased as SL shortened. Peak axial, posterior, and medial ankle contact forces ( $P\leq 0.009$ ) decreased as SL shortened. All pairwise comparisons were significantly different for plantarflexion moment, and axial and posterior forces. All pairwise comparisons except -5%PSL vs. -10%PSL were different for inversion and internal rotation moments and medial forces.

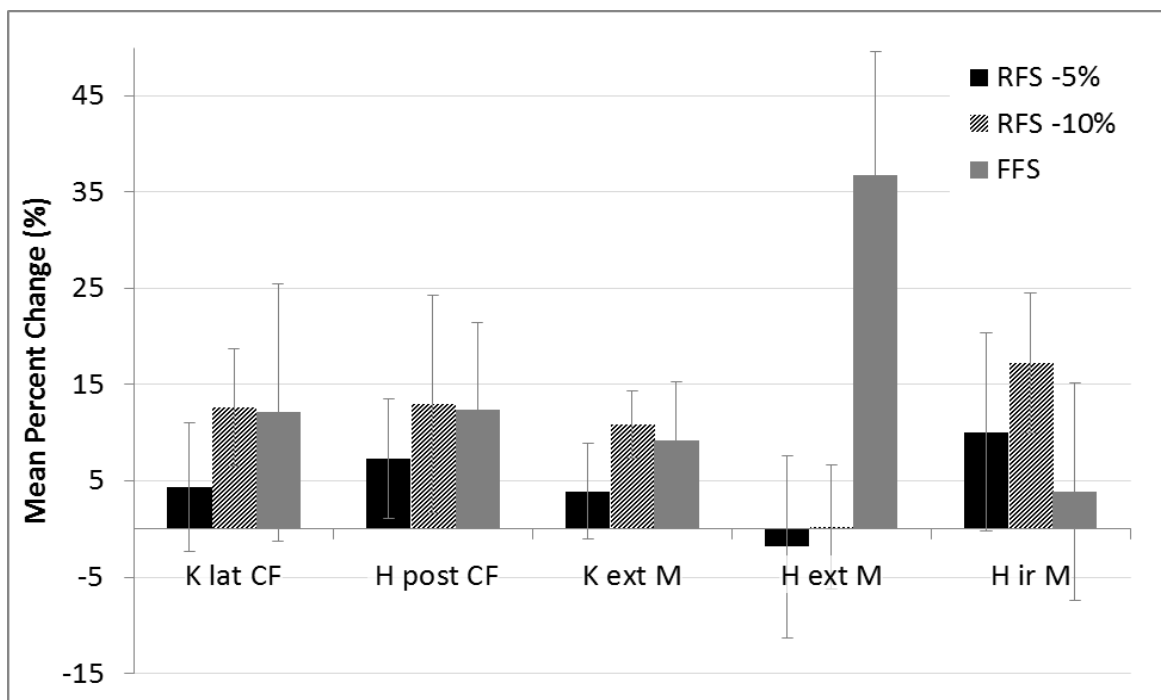
At the knee, peak extension ( $P<0.001$ ), abduction ( $P=0.002$ ), and internal rotation ( $P=0.018$ ) moments decreased with a shorter SL, while external rotation moment remained the same ( $P=0.324$ ). Peak axial, posterior, and lateral contact forces decreased as SL shortened (all

$P \leq 0.001$ ). Peak compressive PFCF and PFS decreased (both  $P < 0.001$ ) with a shorter SL. All pairwise comparisons were significant for extension moment, axial and lateral forces, PFCF, and PFS. Only PSL vs. -10%PSL were different for abduction and internal rotation moments. All but PSL vs. -5%PSL were different for posterior force.

At the hip, peak hip abduction and internal rotation moments decreased (both  $P < 0.001$ ) with a shorter SL, while peak extension moment was unaffected ( $P = 0.273$ ). Surprisingly, peak hip flexion moment increased ( $P = 0.003$ ) as SL shortened. Peak axial, posterior, and lateral contact forces decreased as SL shortened (all  $P < 0.001$ ). All pairwise comparisons were significantly different for all hip contact forces, and abduction and internal rotation moments. All comparisons except -5%PSL vs. -10%PSL were different for hip flexion moment.

#### Shorter Stride Length vs. FFS in hRF

The variables that were elevated for FFS were compared to the 5% and 10% shorter SLs for hRF (Figure 3). Except for hip extension moment, -10%PSL decreased values to the same extent or more than using a FFS.



**Figure 3.** Mean percent change (SD) in variables at the knee (K) and hip (H) for the hRF group using a RFS (-5% and -10%PSL) and a FFS. Positive % change represents decreased loading relative to the reference conditions, PSL for RFS. Variables are ones that were significantly larger during RFS. (ext: extension, ir: internal rotation, post: posterior, lat: lateral, CF=contact force, M=moment)

### Questionnaire

When asked, “Do you think this 5% change is reasonable to maintain during your normal running,” 67% of the hRF and hFF (each 14/21) said yes, stating the shorter SL was not too different or uncomfortable. However, for the -10%PSL condition, only 14% (3/21) of the hRF and 10% (2/21) of the hFF said they could maintain it.

### **Discussion**

Our primary goal of this study was to provide a comprehensive overview of the 3-dimensional lower extremity joint loading differences with a RFS and FFS during shod running with a normal and shorter SL. In general, our data support our hypotheses; loading was greater at the ankle for FFS, some knee variables were greater for RFS, some hip variables were similar for both foot strike styles, and most loads decreased as SL shortened. However, there were some contradictions, namely, several knee variables were greater for FFS, some hip variables were greater for RFS, and peak hip flexion moment increased as SL shortened.

#### Differential response of Groups to Foot Strike Style

Overall, habitual groups were able to replicate the new foot strike style. We are unsure, though, why peak hip flexion moment was largest when habitual groups used the opposite foot strike style.

#### Effects of Foot Strike

##### Ankle

Greater peak axial ankle contact force and plantarflexor moment during FFS agrees with Rooney and Derrick (2013) and is a natural result of greater contraction of the plantarflexors. The greater posterior force for FFS may also be of concern since ultimate shear strength of bone is lower than compressive or tensile strength (Turner et al., 2001). It is possible these larger forces acting at the distal tibia induce greater internal bone stresses, which may propagate into stress fractures if the bone has not adapted to these higher loads. Likewise, the talus may experience increased loads with these axial forces. However, there has been no large-scale empirical evidence to suggest hFF runners are at increased risk of talar or tibial stress fractures, so these hypotheses are speculative. Further modeling and prospective studies may help provide an answer.

Stearne et al. (2014) found similar peak ankle internal rotation moments for habitual groups using their normal foot strike but a large increase when hRF switched to FFS, which our results support. However, we observed hFF significantly decreased this moment when switching to a RFS, which they did not observe. To our knowledge, ankle internal rotation moment has not been associated with running injuries, but Vashishth et al. (2001) did note cycles to failure of bovine tibiae decreased when simultaneous torsional and axial loads were applied, so this may be of concern when using a FFS because of the concurrent increases in torsional moment and axial contact force. We intend to examine this possibility in the future.

### Knee

Despite peak knee extensor moment being 11% greater for RFS, peak axial contact force was approximately 0.8 BW (7%) larger for both groups during FFS. This discrepancy highlights the importance of considering elevated quadriceps activity, which resists knee flexion while the gastrocnemii are active. These results suggest the tibia is under greater axial loading during FFS because of the larger loads at both the ankle and knee. However, the greater lateral shear force may be of concern when using a RFS. Alternatively, both internal and external rotation knee moments were larger for FFS, so both foot strike styles expose the knee to shear stresses. The combined higher torsional moments and axial forces are not tolerated by bones as well, especially when these two types of loads occur more in-phase (George & Vashishth, 2005) as they do for FFS (peak internal rotational moment and axial contact force both occur close to 35-40% of stance). Nevertheless, knee torsional moments are many times smaller than those at the ankle and hip, so they may not be as important for this combined loading. This merits further investigation, but runners should be cautioned that these findings do not support the transition from RFS to FFS to reduce knee or lower leg injuries.

We found that peak knee abduction moment was greater for FFS than RFS, averaged across groups. This was mainly because it drastically increased when hRF used an FFS and may partially be explained by greater peak leg adduction angle (RFS:  $6.6 \pm 1.9^\circ$ , FFS:  $7.6 \pm 1.9^\circ$  for PSL, averaged across groups). Indeed, when using their habitual foot strike style, hRF had a greater knee abduction moment than hFF, which does agree with Kulmala et al. (2013) and Stearne et al. (2014). We also noted a more distinct biomodal curve for FFS than previously reported (Figure 1). Peak knee abduction moment is typically of concern for knee osteoarthritis (Miyazak et al., 2002), but runners do not seem to be at increased risk (Miller et al., 2014).

However, Stefanyshyn et al. (2006) found increased abduction impulse in hRF who developed PF pain, so this variable may be of concern if retraining hRF to use a FFS.

### Patellofemoral

Unlike others (Kulmala et al., 2013; Vannatta & Kernozek, 2015), we found no difference in peak PFCF or PFS between foot strike styles. However, these variables were elevated in hRF versus hFF in a univariate analysis. The lack of differences are attributed to our optimization routines distributing plantarflexion load between the gastrocnemius and soleus approximately equally, whereas others using optimization place approximately 80% of the load in soleus and 20% in gastrocnemius (Lenhart et al., 2014; Vannatta & Kernozek, 2015). Distributing load equally between the two plantarflexor groups is reasonable as Moritani et al. (1991) noted a ratio of approximately 1.0 between the two muscle groups during toe standing and repetitive jumping, which may be partially comparable to midstance of running. Peak PFS was similar to Vannatta and Kernozek (2015) after accounting for speed differences. Since our PF loads appear similar between foot strike styles, hRF who suffer from PF pain syndrome may not benefit by switching to a FFS, unless also shortening their SL.

### Hip

In general, contact forces and moments were larger at the hip for RFS than FFS or they did not differ. With a FFS, the hip essentially extends throughout stance whereas hip flexion occurs after heel contact with a RFS. This requires eccentric hip extensor contraction, which is why peak hip extension moment is larger for RFS and is not mitigated with a shorter. Despite this, the peak axial contact force is not larger for RFS, but posterior contact force is, likely because the flexed orientation of the thigh during that phase of stance. The greater internal rotation moment and posterior contact force may be detrimental to femoral health if unaccustomed to these loads and this may be a reason for hFF runners not to switch to a RFS. Injury data between foot strike groups is lacking, but one study did observe significantly higher incidence of hip pain in hRF versus hFF (who reported no hip pain) (Daoud et al., 2012).

### Effect of Stride Length

Most moments and forces logically decreased as SL shortened, since reduced muscle forces occur because of a shorter SL. The decreased PFCF and PFS may partially be due to decreased peak knee flexion (3.0° or 6%) during -10%PSL, which Lenhart et al. (2014) found to

be the greatest predictor of PFCF. They noted an 11% decrease in PFCF when SL shortened 10%, similar to our 15% reduction.

The smaller frontal plane ankle and hip moments may partially be due to decreased peak flexion angles that occur in these joints during a shorter SL (Boyer et al. 2015; in review). Exceptions to the decreased loading included knee external rotation moment, and hip flexion and extension moments, so injuries related to elevated values for these variables may not benefit from a shorter SL. Unexpectedly, peak hip flexion moment increased as SL shortened. Perhaps there was stronger hip flexor activation just before toe-off to swing the leg forward quicker because of the imposed shorter stride time, supported by greater rectus femoris activation in pre-swing/early swing when increasing step rate 10% (Chumanov et al., 2012).

#### Shorter Stride Length vs. use of FFS in the hRF Group

One of our goals was to see if hRF could use a shorter SL and still lower their loading as much as switching to a FFS. Indeed they could for four of the five variables that were higher with a RFS, notably knee extensor moment. This supports Thompson et al. (2014) who found similar kinetics when running shod or barefoot with the same SL; most decreases associated with barefoot running are typically due to a shorter PSL. Similar precaution must be taken when comparing RFS and FFS since our hRF used a 2.3% shorter PSL during the FFS condition. Using a shorter SL may be a safer alternative for hRF than switching to a different foot strike style that would place additional stresses on maladapted tissues. Hip extensor moment, however, is still greater with a RFS even when shortening SL 10%, so if hip extensor moment is the culprit for injuries in hRF, those runners may try to decrease these loads by using a FFS or another gait modification.

#### Questionnaire

Since two-thirds of runners found it feasible to run with a 5% shorter SL, but less than 15% of runners wanted to use a 10% shorter SL, the larger beneficial decreases in loading at -10%PSL may not be implementable. However, 5% shorter may be feasible but runners will not benefit as much. It is possible the unfavorable responses to the shorter SL conditions may be a result of the majority of our runners already having a reasonable step length. Morgan et al. 1994) defined “overstriders” as runners with step lengths of approximately 138%LL or larger. Our average was 129%LL (range hRF: 118-138%, hFF:118-149%). Interestingly, two of the hFF with the longer relative SLs (138% and 149%) used to be rearfoot strikers.

Interpretation of the results must be considered with the limitations. Static muscle optimization may overestimate muscle forces (Prilutsky et al., 1997) so our magnitudes may be too large, but since the only significant contact force results were repeated measures, it is unlikely the optimization routine differentially overestimated values within SL or foot strike conditions. Similarly, our optimization criteria are not necessarily equivalent to how the human body recruits muscles. For instance, it estimates little to no muscle forces in the hamstrings during midstance (data not shown, Lenhart et al., 2014), but EMG indicates they are active (Chumanov et al., 2012; Giandolini et al., 2013; Shih et al., 2013; Teng & Powers, 2014) and that they are more active in FFS than RFS at midstance (Shih et al., 2013). Moreover, if we allocated greater forces in the soleus and less in the gastrocnemius as others have (Lenhart et al., 2014; Vannatta & Kernozek, 2015), knee and PF contact forces would decrease. Therefore, our relative findings are more meaningful than absolute values. Thirdly, the values for the newly imposed foot strike style represent acute changes. Perhaps runners make additional adaptations that may further increase or decrease loading after regularly using a new foot strike; this should be investigated.

## Conclusions

We report three-dimensional joint loads (contact forces and moments) in the lower extremity for both hRF and hFF with normal and shorter SLs. If runners use a shorter SL, both hRF and hFF will generally decrease their lower extremity joint loads. However, runners may find it more tolerable to use a 5% versus 10% shorter SL. Habitual rearfoot strikers can decrease many joint loads to the same degree associated with switching to a FFS by shortening their SL 10%. In general, FFS increases loads at the ankle and decreases some loads at the knee and hip. The elevated axial and torsional loads at the ankle and knee with a FFS may place the tibia at increased risk for injury. Therefore, FFS and RFS load muscles and bones differently, so one foot strike style is not necessarily worse than the other. It is likely that hRF's and hFF's bones and muscles have adapted or partially adapted to their respectively higher musculoskeletal loads.

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## CHAPTER 5

### SELECT INJURY-RELATED VARIABLES ARE AFFECTED BY STRIDE LENGTH AND FOOT STRIKE STYLE DURING RUNNING

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#### Abstract

Background: Some frontal plane and transverse plane variables have been associated with running injury but it is not known if these variables differ with foot strike style or as stride length is shortened. Purpose: To identify if step width, iliotibial band (ITB) strain and strain rate, positive and negative free moment, pelvic drop, hip adduction, knee internal rotation, and rearfoot eversion differ between habitual rearfoot and habitual midfoot/forefoot strikers when running with both a rearfoot strike (RFS) and midfoot/forefoot strike (FFS) at three different stride lengths. Methods: Healthy runners (21 habitual rearfoot, 21 habitual mid/forefoot) ran overground at 3.35 m/s with both a RFS and FFS at their preferred stride length and 5% and 10% shorter. Results: Variables did not differ between habitual groups. Step width was 1.5 cm narrower for FFS and it widened 0.8 cm as stride length shortened. ITB strain and strain rate did not differ between foot strike but decreased as stride length shortened (0.3% and 1.8%/s, respectively). Pelvic drop was reduced 0.7° for FFS compared to RFS, and both pelvic drop and hip adduction decreased as stride length shortened (0.8° and 1.5°, respectively). Peak knee internal rotation was not affected by foot strike or stride length. Peak rearfoot eversion was not different between foot strike but decreased 0.6° as stride length shortened. Peak positive free moment was not affected by foot strike or stride length. Peak negative free moment was -0.0038 BW·m/ht greater for FFS and decreased -0.0004 BW·m/ht as stride length shortened. Conclusion: The small decreases in most variables as stride length shortened were likely associated with the concomitant wider step width. RFS had slightly greater pelvic drop while FFS had slightly narrower step width and greater negative free moment. Shortening one's stride

length may decrease or at least not increase propensity for running injuries based on the variables we measured. One foot strike style does not appear universally better than the other; rather, different foot strike styles may predispose runners to different types of injuries.

## **Introduction**

Several authors have noted that shortening runners' stride length  $\leq 10\%$  decreases lower extremity loading (Derrick et al., 1998; Derrick et al., 2000; Edwards et al., 2009; Heiderscheit et al., 2011; Lenhart et al., 2014; Willson et al., 2014) and peak hip adduction (Heiderscheit et al., 2011), all while maintaining similar levels of oxygen consumption (Hamill et al., 1995; Cavanagh and Williams, 1982). However, a limitation of these studies is that they only observed the effects of altering stride length in habitual rearfoot strikers (habitual rearfoot) or they did not specify if a rearfoot, midfoot, or forefoot strike was used (Cavanagh & LaFortune, 1980). Therefore, it is unclear if habitual mid-/forefoot strikers (habitual mid/forefoot) also exhibit similar biomechanical changes due to shortened stride length.

Another spatial characteristic of a runner's stride is step width, or the medial-lateral distance between subsequent steps. Typical step width during running is approximately 3-6 cm (Arellano & Kram, 2011; Brindle et al., 2014; Kulmala et al., 2013; Meardon et al., 2012) and does not seem to differ between habitual foot strike groups (Kulmala et al., 2013). A narrower step causes greater iliotibial band (ITB) strain and strain rate (Meardon et al., 2012), increased frontal plane ankle and knee moments (Brindle et al., 2014), and increased peak rearfoot eversion and hip adduction angles (Brindle et al., 2014). Greater hip adduction and knee internal rotation have been retrospectively and prospectively found in runners with ITB syndrome and patellofemoral pain syndrome (Ferber et al., 2010; Noehren & Davis, 2007; Noehren et al., 2007; Wilson & Davis, 2008), and peak hip adduction and rearfoot eversion are greater in females with a history of tibial stress fractures (Milner et al., 2010). Heiderscheit et al. (2011) noted that peak hip adduction angle decreased as step rate increased, which may suggest runners concomitantly widened their step width as they shortened their stride, but this has yet to be assessed. Finally, greater contralateral pelvic drop (likely due to weaker hip abductors) is also associated with ITB syndrome and patellofemoral pain syndrome (Dierks et al., 2008; Fredericson et al., 2000; Wilson and Davis, 2008), so it is noteworthy to look at all the

aforementioned variables in more detail as we systematically alter stride length and foot strike style, two common clinical or coaching manipulations.

A narrower step and greater pronation have been associated with a larger free moment (Holden & Cavanagh, 1991), which is the reaction moment to the torsional moment applied by the foot on the ground (Holden & Cavanagh, 1991). A shod mid-/forefoot strike (FFS) exhibits greater eversion excursion compared to rearfoot strike (RFS) because the foot is more inverted at contact (Laughton et al., 2003; Stackhouse et al., 2004). Because of these differences and the coupling between pronation and tibial rotation (Lundberg, 1989), free moment likely differs between foot strike styles. Quantifying free moment between foot strike styles and during shorter stride lengths has not been done but is important as some tibial stress fractures appear torsional in nature (Spector et al., 1983) and greater free moment is retrospectively and prospectively associated with tibial stress fracture history (Milner et al., 2006; Pohl et al., 2007; Pohl et al., 2008).

All articles mentioned thus far reported neutral or positive changes in oxygen consumption, kinematics, and kinetics while shortening or widening the stride. However, all of these studies have only looked at habitual rearfoot (Derrick et al., 1998; Derrick et al., 2000; Heiderscheit et al., 2011; Holden & Cavanagh, 1991; Milner et al., 2006; Pohl et al., 2008), did not state the foot strike pattern used by runners (Arellano & Kram, 2011; Brindle et al., 2014; Edwards et al., 2008; Lenhart et al., 2014; Meardon & Derrick, 2008;), or did not differentiate between foot strike style groups (26). These variables should be investigated in FFS, especially with a recent interest in minimalist shoes and barefoot running, which generally encourages the use of a FFS (De Wit et al., 2000; Hamill et al., 2011; Lieberman et al., 2010). Therefore, the primary purpose of this study was to compare step width, free moment, ITB strain and strain rate, and select lower extremity frontal and transverse plane kinematics when stride length was shortened 5% and 10% in both habitual rearfoot and habitual mid/forefoot using both foot strike patterns while shod. It was hypothesized that step width would widen as stride length shortened, which would be accompanied by decreased peak hip adduction, peak rearfoot eversion, and peak ITB strain and strain rate. Our second hypothesis was that both peak positive and negative free moment would decrease as stride length shortened, assuming a concomitant wider step. Finally, we hypothesized step width and free moment would be similar for both foot strike styles.

## Methods

Forty-two recreational or competitive runners (21 habitual rearfoot and 21 habitual mid/forefoot) volunteered. Groups were similar, except the habitual rearfoot were taller, reflecting an unequal number of women (1 in habitual rearfoot, 7 in habitual mid/forefoot) (habitual rearfoot age:  $21 \pm 6$  yrs; mass:  $72.1 \pm 10.3$  kg; height:  $1.81 \pm 0.07$  m; leg length:  $0.965 \pm 0.039$  m; weekly mileage:  $30.0 \pm 21.2$  mi) (habitual mid/forefoot age:  $22 \pm 4$  yrs; mass:  $66.7 \pm 10.5$  kg; height:  $1.74 \pm 0.09$  m; leg length:  $0.921 \pm 0.057$  m; weekly mileage:  $35.1 \pm 27.6$  mi). An analysis with sex as the independent variable did not yield any significant effects for any variables (all  $P \geq 0.160$ ), so unequal numbers of women in the groups is not believed to have affected our results. We grouped midfoot and forefoot strikers and strikes together as habitual mid/forefoot or FFS because of similar sagittal ankle kinematics and GRFs observed in our runners and to increase statistical power because of the relative scarcity of forefoot strikers (Cavanagh & LaFortune, 1980). The study was approved by the university's Institutional Review Board, and written informed consent was obtained prior to participation.

Participants visited the lab once, during which 25 individual reflective markers were placed on the lower extremity and shoes (sacrum, and bilaterally on the ASIS, greater trochanter, lateral thigh, anterior distal thigh, lateral femoral condyle, anterior proximal and distal leg, posterior leg, lateral malleolus, dorsifoot, lateral foot, and heel). Additional markers were placed on the medial femoral condyles and malleoli for the standing calibration trial used to identify knee and ankle joint centers. Participants performed all conditions in their own shoes (none wore barefoot-simulating footwear). First, they warmed up at a self-selected speed on the treadmill and then ran 5 minutes at 3.35 m/s with their habitual foot strike. Time to complete 20 strides was timed after minute one and minute three of the 5 minute run, and the average was found. This was used to determine preferred stride length and step rate. No other data were collected during treadmill running.

Nine tape marks were placed on the floor, perpendicular to the running path, at distances corresponding to the runner's preferred step length. Pieces of athletic tape, approximately 19 cm long with an approximately 6.4 cm long marker line in the middle, were used. Participants performed seven overground running trials at 3.35 m/s ( $\pm 3\%$ ) after completing the treadmill running, landing with their right foot on the force platform. Next, participants ran on the treadmill again for three minutes in sync with a metronome indicating either a 5% or 10% shorter



stride length. This was followed by the overground trials with the shortened stride length, while running at 3.35 m/s. This process was repeated for the other shortened stride length condition. The order of presentation of the shortened stride length conditions was counter balanced.

Brief instructions were then given on how to run with the opposite strike pattern, including a few seconds of slow-motion video showing the correct foot strike on a treadmill. The three experimental conditions were repeated but with the opposite foot strike style. A new preferred stride length was found for the new foot strike style.

Strike index was calculated as the average center of pressure location during the first 10 ms of stance and reported as a percentage of foot length from the posterior calcaneus. Foot strikes were categorized as RFS (strike index < 33.3%) or FFS (strike index > 33.3%) and were verified after each overground trial. If a runner did not use the correct foot strike style, did not run at the correct speed, or missed placement of the foot on the force platform (i.e., not within  $\pm 2.5\%$  of the target stride length), a new trial was collected.

Kinematic data were collected at 200 Hz with an 8-camera Vicon system (Vicon MX, Vicon, Centennial, CO, USA) and low-pass filtered at 20 Hz. Kinetic data were collected at 1600 Hz by an in-ground AMTI force platform (AMTI, Watertown, MA) and were filtered at 50 Hz. Cardan angles were calculated for the lower extremity segments and joints using a 'zxy' rotation order (flexion/extension, ab/adduction, internal/external rotation). Stride length was calculated as the anterior-posterior (AP) distance between the left heel markers for the steps before and after the force platform and is reported as a percent of leg length due to differences in height between groups. Leg length was defined as the distance from the superior border of the greater trochanter to the floor. Stride length and step width were calculated using the average heel marker position while its AP velocity was near zero ( $< 0.15$  m/s) during stance. Step width was calculated as the mediolateral distance from the right heel marker during stance to a virtual line connecting the left heel markers for the steps before and after the force platform. Free moment was calculated according to Holden and Cavanagh (1991) and was normalized to body weight and height, giving units of BW·m/ht. A positive free moment indicates an adductor free moment, which resists a net toeing out moment, while a negative free moment indicates an abductor free moment, which resists a net toeing in moment.

ITB strain and strain rate were calculated using a scaled lower extremity model from OpenSIM (2), except we defined the ITB attachments using the five points from Meardon et al.

(26). Animation of the model was implemented using custom Matlab software (7.13.0.564, R2011b, Natick, MA, USA). ITB length was calculated by summing the individual segment lengths of the tensor fascia lata muscle/tendon. ITB strain was calculated using the formula from Hamill et al. (2008):

$$\text{Strain} = \frac{L_i - L}{L} \times 100$$

where  $L_i$  is ITB length during time  $i$  of the stance phase, and  $L$  is the ITB length during the standing trial. ITB strain rate was calculated using the first central difference method of the strain data.

A  $2 \times 2 \times 3$  repeated-measures MANOVA (habitual group  $\times$  foot strike  $\times$  stride length) was performed in SPSS to compare step width, peak adductor free moment and abductor free moment in the first half of stance, peak hip adduction, peak contralateral pelvic drop, peak knee internal rotation, peak ankle eversion angle, peak ITB strain, and peak ITB strain rate. Significance was set to  $\alpha=0.05$ . If the MANOVA was significant, the univariate ANOVAs were subsequently evaluated at a 0.05 level. If the assumption of sphericity was violated, Greenhouse-Geisser corrections were used. Polynomial contrasts were used to identify significant trends for stride length. Last value carried forward was used for a few conditions with missing data. Transverse plane kinematics were assessed post-hoc to help explain differences in free moment.

## Results

Participants successfully replicated the alternate foot strike style (strike index for preferred stride length RFS, habitual rearfoot:  $2 \pm 2\%$  and habitual mid/forefoot:  $2 \pm 2\%$ ,  $P = 0.653$ ; for preferred stride length FFS, habitual rearfoot:  $56 \pm 6\%$  and habitual mid/forefoot:  $58 \pm 7\%$ ,  $P = 0.268$ ). They were also relatively successful at shortening their stride the prescribed amount, although they tended to not shorten their stride the full amount (actual stride length reductions: 4.4% and 8.9%, averaged across group and foot strike style). Preferred stride length was similar for both groups using their normal foot strike style (habitual rearfoot:  $258.5 \pm 12.3\%$  of leg length, habitual mid/forefoot:  $256.8 \pm 15.2\%$  of leg length,  $P = 0.696$ ). Habitual rearfoot strikers shortened their stride by  $5.5 \pm 8.5$  cm (2.2%) when switching to a FFS ( $P = 0.007$ ), although four habitual rearfoot lengthened their stride length. habitual mid/forefoot did not change their preferred stride length (FFS:  $256.8 \pm 15.2\%$  of leg length; RFS:  $258.4 \pm 15.8\%$  of leg length,  $P = 0.167$ ) when asked to run with a RFS.

There were no significant three-way ( $P = 0.408$ ) or two-way interactions (group $\times$ foot strike  $P = 0.189$ ; group $\times$ stride length  $P = 0.450$ ; foot strike $\times$ stride length  $P = 0.280$ ). The within-subject main effects were significant (foot strike style:  $P < 0.001$ ; stride length:  $P < 0.001$ ) but the between-subjects effect was not (habitual group:  $P = 0.215$ ). Therefore, we only explored the univariate tests for foot strike and stride length, collapsed across the two habitual groups. Means and standard deviations for all dependent variables are presented in Tables 1 and 2.

Step width was different between foot strike and stride length conditions. Step width was approximately 1.5 cm wider in RFS versus FFS ( $0.0 \pm 3.1$  and  $-1.5 \pm 3.9$  cm,  $\eta_p^2 = 0.406$ ,  $P < 0.001$ ) and became wider as stride length shortened ( $-1.1 \pm 3.9$  to  $-0.8 \pm 3.6$  to  $-0.3 \pm 3.4$  cm for preferred stride length, 5%, and 10% shorter, respectively;  $\eta_p^2 = 0.334$ , linear trend  $P < 0.001$ ,  $y = 0.084x - 1.2$ ,  $r^2 = 0.996$ ). We compared transverse foot segment orientation at contact for the preferred stride length conditions to see if it was affecting our step width calculation and found with a FFS, the foot was more abducted (RFS:  $-5.7 \pm 4.8^\circ$ , FFS:  $-8.8 \pm 4.9^\circ$ ). Accounting for this by adjusting FFS foot orientation to be similar to RFS by rotating about the ankle joint center, Step width for FFS widened by 0.7 cm. Even with this adjustment, step width was still narrower for FFS (preferred stride length averaged across groups, RFS:  $-0.1 \pm 3.2$  cm, FFS:  $-1.4 \pm 4.1$  cm,  $P = 0.003$ ).

**Table 1.** Mean (SD) for FM, and kinematic variables for both groups for all conditions. hRF: habitual rearfoot strikers, hFF: habitual mid/forefoot strikers, RFS: rearfoot strike, FFS: mid/forefoot strike, PSL: preferred stride length, FM: free moment

	Peak adductor FM (BW·m/ht) x10 <sup>-3</sup>			Peak abductor FM (BW·m/ht) x10 <sup>-3</sup> *,^			Contralateral pelvic drop (°)*,^			Peak hip adduction (°)^			Peak knee internal rotation (°)			Peak ankle eversion (°)^		
	PSL	5%	10%	PSL	5%	10%	PSL	5%	10%	PSL	5%	10%	PSL	5%	10%	PSL	5%	10%
<b>hRF</b>																		
<b>RFS</b>	4.2 (2.5)	5.0 (2.8)	4.7 (2.4)	-1.5 (1.1)	-1.2 (1.2)	-0.9 (1.0)	-5.6 (3.1)	-5.5 (2.8)	-5.1 (2.9)	17.4 (4.6)	17.1 (4.3)	16.3 (4.3)	13.2 (3.8)	13.5 (3.7)	13.4 (3.8)	-12.3 (7.8)	-12.2 (7.9)	-12.1 (7.9)
<b>FFS</b>	4.9 (2.4)	5.0 (2.0)	4.8 (2.5)	-5.1 (2.0)	-4.9 (1.8)	-4.6 (1.6)	-5.5 (2.9)	-5.1 (2.7)	-4.6 (2.8)	17.6 (4.6)	16.9 (4.1)	15.6 (4.3)	12.9 (4.8)	13.1 (4.7)	13.1 (5.0)	-12.7 (7.5)	-12.5 (8.4)	-12.2 (8.3)
<b>hFF</b>																		
<b>RFS</b>	5.3 (2.6)	5.4 (2.4)	5.5 (2.4)	-1.1 (1.0)	-0.8 (0.9)	-0.8 (0.9)	-6.6 (2.2)	-6.2 (1.8)	-5.5 (2.2)	16.6 (3.7)	16.2 (3.4)	15.3 (3.2)	13.1 (8.1)	13.2 (7.3)	12.7 (7.2)	-9.9 (7.0)	-9.5 (6.8)	-9.8 (7.1)
<b>FFS</b>	6.0 (3.2)	6.1 (3.3)	5.5 (3.3)	-5.0 (1.1)	-4.9 (1.1)	-4.7 (1.3)	-5.4 (1.9)	-5.1 (1.9)	-4.6 (1.9)	16.1 (3.3)	15.3 (3.2)	14.5 (3.4)	11.8 (6.6)	12.2 (6.5)	12.3 (6.6)	-11.4 (6.9)	-10.3 (7.1)	-9.8 (7.0)

\* Significantly different between foot strike styles, averaged across groups and SL conditions (p<0.05)

^ Significant linear trend for stride length, averaged across groups and foot strike style (p<0.05)

**Table 2.** Mean (SD) for step width and ITB variables for both groups for all conditions. hRF: habitual rearfoot strikers, hFF: habitual mid/forefoot strikers, RFS: rearfoot strike, FFS: mid/forefoot strike, PSL: preferred stride length, ITB: iliotibial band

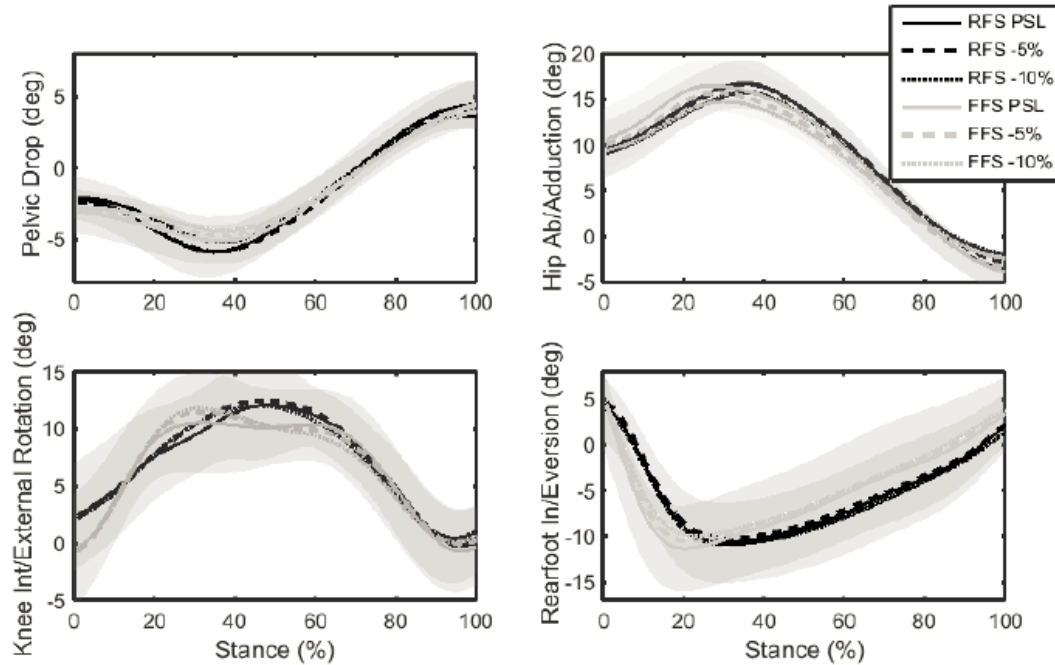
	Step width (cm)*,^			ITB strain (%)^			ITB strain rate (%/s)^		
	PSL	5%	10%	PSL	5%	10%	PSL	5%	10%
<b>hRF</b>									
<b>RFS</b>	-1.1 (3.7)	-0.9 (3.6)	-0.3 (3.3)	4.7 (0.5)	4.7 (0.4)	4.6 (0.4)	39.4 (12.9)	39.8 (13.0)	38.8 (13.3)
<b>FFS</b>	-2.7 (5.2)	-2.6 (4.3)	-1.7 (4.2)	5.0 (0.6)	4.8 (0.5)	4.6 (0.5)	44.9 (12.9)	44.0 (14.7)	40.2 (11.9)
<b>hFF</b>									
<b>RFS</b>	0.8 (2.4)	0.9 (2.5)	0.9 (2.7)	4.6 (0.6)	4.6 (0.5)	4.4 (0.6)	37.4 (11.6)	37.8 (11.3)	37.7 (11.0)
<b>FFS</b>	-1.6 (3.1)	-0.4 (2.9)	0.0 (2.9)	4.7 (0.6)	4.5 (0.6)	4.4 (0.6)	38.6 (13.1)	38.4 (13.3)	36.1 (12.9)

\* Significantly different between foot strike styles, averaged across groups and SL conditions (p<0.05)

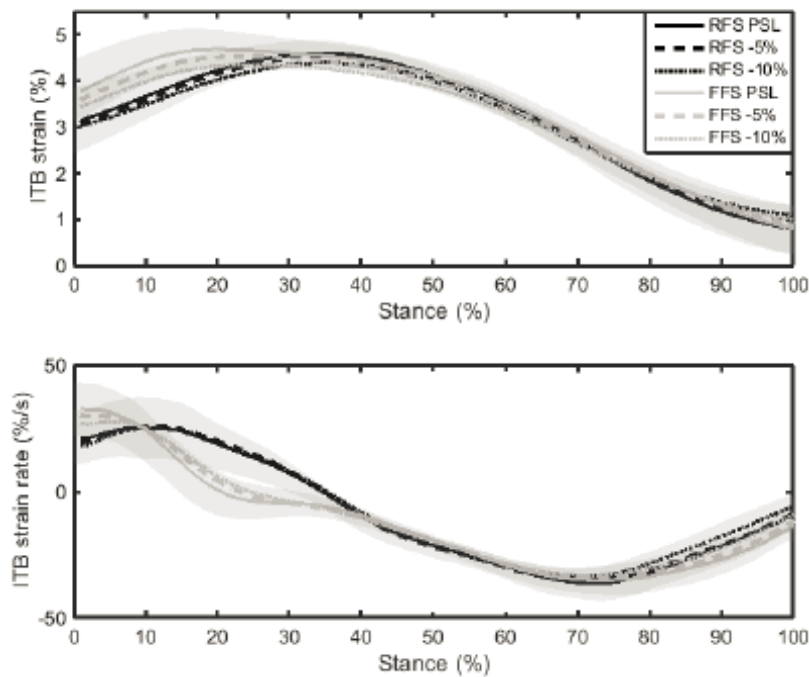
^ Significant linear trend for stride length, averaged across groups and foot strike style (p<0.05)

Kinematics were affected by foot strike style and stride length (Figure 1). Pelvic drop was slightly greater for RFS than FFS ( $-5.8 \pm 2.6^\circ$  versus  $-5.1 \pm 2.4^\circ$ ;  $\eta_p^2 = 0.295$ ,  $P < 0.001$ ) and decreased as stride length shortened ( $-5.8 \pm 2.6^\circ$ ,  $-5.5 \pm 2.3^\circ$ ,  $-5.0 \pm 2.5^\circ$ ;  $\eta_p^2 = 0.635$ , linear trend  $P < 0.001$ ,  $y = 0.08x - 5.8$ ,  $r^2 = 0.976$ ). Peak hip adduction angle was similar between foot strike styles (RFS:  $16.5 \pm 3.9^\circ$ , FFS:  $16.0 \pm 3.9^\circ$ ;  $\eta_p^2 = 0.040$ ,  $P = 0.075$ ) and it decreased as stride length shortened ( $16.9 \pm 4.1^\circ$ ,  $16.4 \pm 3.8^\circ$ ,  $15.4 \pm 3.8^\circ$ ,  $\eta_p^2 = 0.775$ , linear trend  $P < 0.001$ ,  $y = -0.14x + 17.0$ ,  $r^2 = 0.975$ ). Although significant, these differences were approximately  $1^\circ$  or less. Peak knee internal rotation angle was similar between foot strike styles (RFS:  $13.2 \pm 5.9^\circ$ , FFS:  $12.6 \pm 5.6^\circ$ ,  $\eta_p^2 = 0.110$ ,  $P = 0.116$ ) and stride length ( $12.7 \pm 6.0^\circ$ ,  $13.0 \pm 5.6^\circ$ ,  $12.9 \pm 5.7^\circ$ ,  $\eta_p^2 = 0.005$ , linear trend  $P = 0.114$ ). Peak eversion angle was similar between foot strike styles (RFS:  $-11.0 \pm 7.4^\circ$ , FFS:  $-11.5 \pm 7.5^\circ$ ,  $\eta_p^2 = 0.060$ ,  $P = 0.295$ ), but as stride length shortened, peak eversion angle decreased ( $-11.6 \pm 7.3^\circ$ ,  $-11.1 \pm 7.6^\circ$ ,  $-11.0 \pm 7.5^\circ$ ,  $\eta_p^2 = 0.196$ , linear trend  $P = 0.005$ ,  $y = 0.057x - 11.5$ ,  $r^2 = 0.914$ ).

Peak ITB strains were similar between foot strike styles (RFS:  $4.6 \pm 0.5\%$ , FFS:  $4.7 \pm 0.6\%$ ,  $\eta_p^2 = 0.025$ ,  $P = 0.147$ ) but decreased as stride length shortened ( $4.8 \pm 0.6\%$ ,  $4.7 \pm 0.5\%$ ,  $4.5 \pm 0.5\%$ ;  $\eta_p^2 = 0.668$ , linear trend  $P < 0.001$ ,  $y = -0.024x + 4.771$ ,  $r^2 = 0.994$ ). Peak ITB strain rate was also similar between foot strike styles (RFS:  $38.5 \pm 12.0\%/s$ , FFS:  $40.4 \pm 13.3\%/s$ ,  $\eta_p^2 = 0.008$ ,  $P = 0.261$ ) but decreased as stride length shortened ( $40.0 \pm 12.7\%/s$ ,  $40.0 \pm 13.1\%/s$ ,  $38.2 \pm 12.2\%/s$ ,  $\eta_p^2 = 0.039$ , linear trend  $P = 0.043$ ,  $y = -0.16x + 40.3$ ,  $r^2 = 0.784$ ). Although ITB strain and strain rate were similar between foot strike styles, the peaks occurred earlier in stance for FFS (Figure 2).

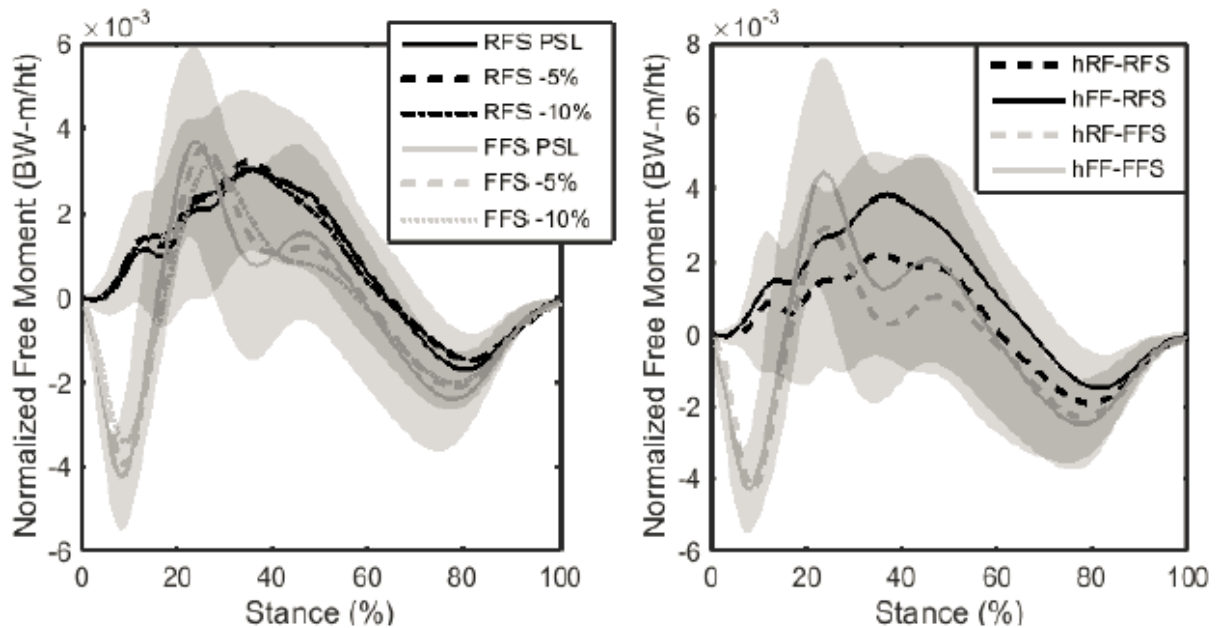


**Figure 1.** Ensemble curves for kinematic variables for the six conditions, averaged across habitual group. RFS conditions are in black, FFS conditions are in gray. Positive values represent contralateral pelvic rise, hip adduction, knee internal rotation, and rearfoot inversion. Shaded gray regions represent 1 standard deviation. PSL: preferred stride length

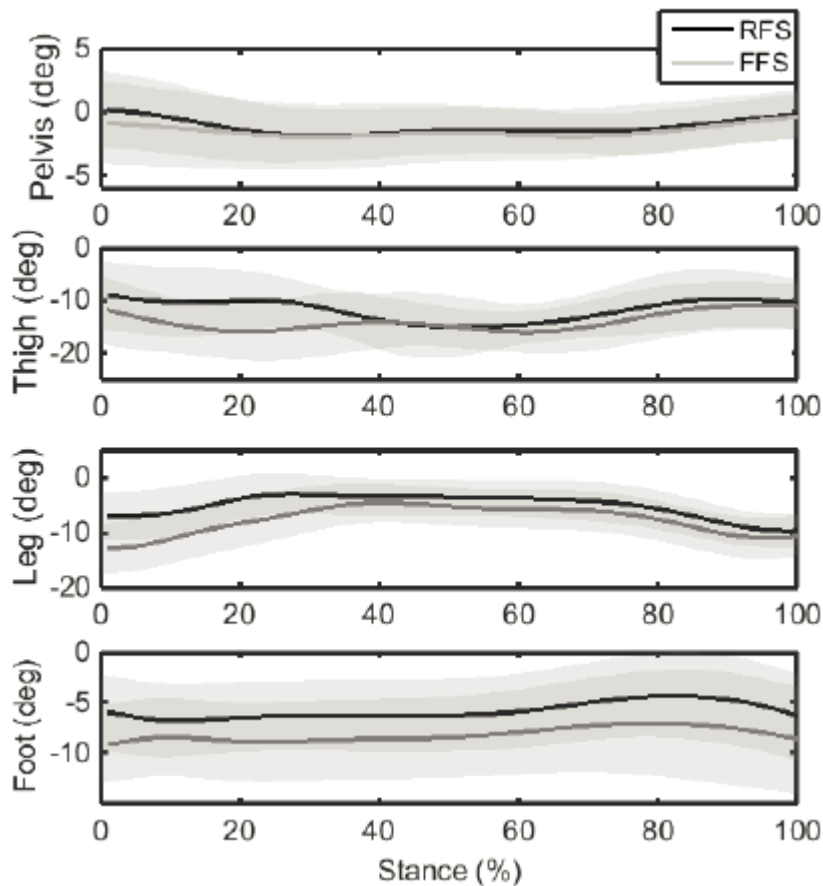


**Figure 2.** Ensemble curves for ITB strain (top) and strain rate (bottom) for the six conditions, averaged across habitual group. RFS conditions are in black, FFS conditions are in gray. Shaded gray regions represent 1 standard deviation. PSL: preferred stride length

Figure 3 shows ensemble curves for normalized free moment. Peak adductor free moment in the first half of stance was similar between foot strike styles (RFS:  $0.0050 \pm 0.0025$ , FS:  $0.0054 \pm 0.0028$  BW·m/ht,  $\eta_p^2 = 0.031$ ,  $P = 0.160$ ) and did not differ as stride length shortened ( $0.0051 \pm 0.0027$ ,  $0.0054 \pm 0.0026$ ,  $0.0051 \pm 0.0026$  BW·m/ht,;  $\eta_p^2 = 0.004$ ,  $P = 0.092$ ). Peak abductor free moment in the first half of stance was affected by foot strike style and stride length. It was over three times larger in FFS versus RFS ( $-0.0049 \pm 0.0096$  versus  $-0.0011 \pm 0.0014$  BW·m/ht,  $\eta_p^2 = 0.823$ ,  $P < 0.001$ ), and it decreased as stride length shortened ( $-0.0032 \pm 0.0023$ ,  $-0.0030 \pm 0.0023$ ,  $-0.0028 \pm 0.0022$  BW·m/ht;  $\eta_p^2 = 0.131$ , linear trend  $P < 0.001$ ,  $y = 0.00004x - 0.0032$ ,  $r^2 = 0.990$ ).



**Figure 3.** Ensemble curves for normalized free moment for the six conditions (left), averaged across habitual group, and for preferred stride length for the habitual foot strike groups running with both foot strike patterns (right). RFS conditions are in black. FFS conditions are in gray. Shaded gray regions represent 1 standard deviation. PSL: preferred stride length, hRF: habitual rearfoot strikers, hFF: habitual mid/forefoot strikers.



**Figure 4.** Ensemble curves for segment internal (positive) and external (negative) rotations, averaged across habitual group and stride length condition. RFS is in black. FFS is in gray. Shaded gray regions represent 1 standard deviation.

## Discussion

The aim of this study was to note how step width, free moment, ITB strain and strain rate, and select frontal and transverse plane kinematics differed as stride length was shortened during both RFS and FFS in both habitual rearfoot and habitual mid/forefoot strikers. Since none of the measured variables differed between habitual foot strike group, changes imposed by shortening one's stride or using a different foot strike can be generalized across groups. Additionally, no foot strike  $\times$  stride length interaction indicates the effects of stride length generalize to both RFS and FFS.

### Effects of Stride Length

Our hypothesis that step width would be wider with a shorter stride length was supported. We observed a significant linear trend toward a wider step width as stride length shortened,



regardless of group or foot strike style being used. This is the first study to identify this phenomenon. Perhaps as participants took quicker steps to shorten their stride length, they had insufficient time to bring their foot through and around the stance limb. Since a wider step decreases ITB strain, joint moments, frontal plane motion, and free moment (Meardon & Derrick, 2008; Meardon et al., 2012 & Brindle et al., 2014), this appears to be an unanticipated benefit for runners who shorten their stride length. However, the change from preferred stride length to -10% preferred stride length ( $< 1.7$  cm) was not nearly as much as the imposed step width changes in previous research ( $\geq 5$  cm) (Meardon et al., 2012 & Brindle et al., 2014), so our participants may not have experienced clinically significant decreases in relevant variables.

Accompanying a wider step width as stride length shortened was decreased pelvic drop, peak hip adduction, peak ankle eversion, peak ITB strain and strain rate, and peak abductor free moment, consistent with previous research (Brindle et al., 2014; Heiderscheit et al., 2011; Meardon & Derrick, 2008; Meardon et al., 2012). These results support our hypotheses. These frontal plane kinematic changes with a shortened stride length may be a result of a wider step, although to be certain, one would need to shorten stride length while holding step width constant. We did not observe a decrease in adductor free moment as hypothesized, suggesting either no change in upper or lower body external rotational moment, or any changes were canceled out by the other half. While we observed statistically significant decreases in the kinematic variables, these decreases may not be large enough to have clinical importance because changes were approximately  $< 1^\circ$ . For instance, ITB syndrome and patellofemoral pain syndrome are associated with about  $2.5$ - $4.0^\circ$  more hip adduction (Ferber et al., 2010; Noehren et al., 2007; ; Noehren & Davis, 2007; Wilson & Davis, 2008) and 0.9% greater ITB strains and 12.7%/s greater ITB strain rates relative to controls (Hamill et al, 2008). As such, shortening one's stride length may not alleviate pain associated with ITB syndrome or patellofemoral pain syndrome, but it should at least not aggravate it.

#### Effects of Foot Strike Style

The different foot strike styles affected step width, peak abductor free moment, and peak contralateral pelvic drop. First, step width was wider for RFS compared to FFS, which fails to support our hypothesis. Despite a narrower step width during FFS, it did not increase peak ITB strain or strain rate. Therefore, it does not appear that using a RFS instead of a FFS would increase risk of ITB symptoms caused by excessive strain or strain rate, lending support for no

differences in rates of ITB syndrome in habitual rearfoot versus habitual mid/forefoot (Dauod et al., 2012). The lack of differences may partly be because transverse foot angle in FFS resulted in a 0.7 cm narrower step width as calculated with heel markers. This emphasizes the use of the ankle joint center, or another landmark that is less susceptible to transverse foot rotations, to calculate step width. We attempted to use ankle joint center, but some foot markers were out of the field of view for the step after the force platform. It should be noted that our average step width for most conditions was a crossover step—much narrower than previously reported (Arellano & Kram, 2011; Brindle et al., 2014; Kulmala et al., 2013; Meardon et al., 2012). This may result from slightly different methods of controlling and calculating step width (e.g., using tape parallel to the running path (Brindle et al., 2014), calculating step width at the instant of contact (Brindle et al., 2014), calculating step width when the heel marker was at its minimum height (Arellano & Kram, 2011, Meardon et al., 2012)). It may also be due to data from Arellano and Kram (2011) being calculated during treadmill running, or a couple of our participants having very narrow step width or participants targeting the tape marks. The same tape marks were used for an entire data collection, so it is possible all values may be biased to be narrower than previously reported if participants targeted foot placement, but the relative effects of foot strike and stride length are valid. Since there were no between group differences, this limitation does not impact our trends.

Peak hip adduction angle was similar between RFS and FFS, which supports similar ITB syndrome and patellofemoral pain syndrome injury rates in habitual rearfoot and habitual mid/forefoot (Dauod et al., 2012). Peak knee internal rotation was similar between foot strike styles, suggesting any pathologies associated specifically with this variable may not be affected by runners' foot strike style. Peak eversion angle was similar between foot strike styles, contradicting Stackhouse et al. (2004) who found decreased peak eversion (but increased excursion) with a FFS versus RFS. Stackhouse et al. (2004) had habitual rearfoot run with both foot strikes rather than using habitual mid/forefoot, as our study did, but both of our groups tended to have increased peak eversion with a FFS.

Lastly, free moment differed between foot strike styles, which has not been reported previously. Even though free moment is quite variable (Holden & Cavanagh, 1991; Milner et al., 2006; Pohl et al., 2008), the pattern for FFS was more consistent than RFS. We speculate the difference in the initial part of stance for FFS may be due to upper body rotation (not measured)

or foot rotation. During RFS, the foot, thigh, and pelvis are externally rotating in the early part of stance (Figure 4), which would create an opposing adductor free moment, while only the shank internally rotates initially. Because of the larger mass of the thigh and pelvis, presumably free moment is more reflective of their motions. For FFS, all transverse motion is similar to RFS, except that *both* the foot and shank internally rotate, and the timing of internal foot rotation and abductor free moment are similar, peaking around 9% of stance. Perhaps this and/or upper body rotation (not measured) resulted in the initial abductor free moment for FFS.

Our hypothesis that adductor free moment would be larger for FFS was not supported. The later occurrence of peak adductor free moment around midstance during RFS (when vertical GRF and muscle forces peak) may be of concern since combined in-phase axial and torsional loads decrease cortical bone fatigue life about 7 times relative to 90° out-of-phase combined loading (George & Vashishth, 2005). This should be explored in more detail. Additionally, the association of rate of free moment loading and running injuries has not been investigated, to our knowledge, and it is clearly larger in both directions for FFS.

A few limitations of the study should be considered. We allowed runners to use their own shoes as it represents what they use outside the lab (none solely ran in barefoot-simulating footwear), but this may have affected between-group differences. Second, runners received minimal instruction when switching to the new foot strike style, so it may not reflect how runners would run after prolonged practice. Third, these were healthy runners at the time of collection, so results may not apply to injured populations. Fourth, several variables reported and injuries associated with abnormal levels of these variables are different between sex, and our groups were not balanced on sex (although statistically it did not affect our results). Fifth, our sample size limited our ability to detect differences between groups since between-subject variability is quite large for most of the variables. For instance, we would have needed approximately 150 runners per group to detect differences in peak hip adduction. Sixth, using the AP velocity of a forefoot (instead of heel) marker may have been more appropriate for FFS to identify foot contact. Additionally, when modeling the ITB, we only considered passive strain. Lastly, our habitual mid/forefoot strikers could have converted from a RFS to a FFS or have always run with a FFS. More distinct group or foot strike differences may have been observed if only natural born habitual mid/forefoot runners were included, which should be addressed in future studies.

## Conclusions

A modestly shorter stride length resulted in a slightly wider step, which was accompanied by small decreases in pelvic drop, peak hip adduction, ankle eversion, ITB strain and strain rate, and peak abductor free moment. This was true for both RFS and FFS and regardless of runners' habitual foot strike style. Step width was narrower and abductor free moment was larger during FFS versus RFS, while pelvic drop was greater with a RFS. Whether such small differences in these variables between foot strike styles predispose runners to sustaining different types of injuries should be explored.

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## CHAPTER 6

### MODIFICATION OF FOOT STRIKE PATTERN INCREASES GAIT CYCLE COMPLEXITY IN RUNNERS

A paper to be submitted to

*Gait and Posture*

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#### Abstract

Non-preferred movements appear to alter the neuromotor timing of cyclical tasks, such as running. Some authors argue that if one's movement is too predictable or too unpredictable, the mover is not in a state of optimal adaptability, which may lead to injury. Our purpose was to see if runners performing novel foot strike and barefoot running would affect the randomness/predictability of their stride time. To quantify this, we used the scaling coefficient alpha ( $\alpha$ ) from detrended fluctuation analysis. Seventeen habitual rearfoot and 16 habitual mid/forefoot strikers participated. They performed three, 8-minute conditions on the treadmill at their self-selected speed: 1) shod running with their normal foot strike style, 2) shod running with their non-preferred foot strike style, and 3) barefoot running. An accelerometer was adhered to their tibia to measure accelerations and subsequently calculate stride time.  $\alpha$  significantly increased for both groups of runners when they performed the novel foot strike conditions, although not all participants responded the same. A larger  $\alpha$  may indicate that the runner less adaptable, possibly increasing their risk of injury.

#### Introduction

Inconspicuous information lies within the several hundred or thousand stride intervals of continuous locomotion. Hausdorff et al. (1995) first discovered this fractal-like characteristic in



human gait. If within a large time-series, such as heart beats or stride times, one value depends on the value of previous points, long range correlations are said to exist. The series demonstrates self-similarity or predictability. Several researchers have looked at this time-dependent repeatability of gait in different diseases and populations. With Parkinson's Disease, Huntington's Disease, amyotrophic lateral sclerosis, elderly fallers, and ACL reconstruction, the time-dependent repeatability was more random than healthy controls (Costa et al., 2003; Hausdorff et al., 1998; Hausdorff, Mitchell et al. 1997; Frenkel-Toledo et al., 2005; Hausdorff, Lertratanakul et al., 2000; Moraiti et al., 2010; Herman et al., 2005). Some researchers believe this is unfavorable because the mover is thought to be most adaptable to task or environmental constraints if there is more patterning/complexity (Stergiou & Decker, 2011). Interestingly, peripheral neuropathy (peripheral sensory nerves lose function but supraspinal centers do not) does not affect long range correlations (Gates & Dingwell, 2007). Therefore, the underlying complexity is likely controlled in the brain stem, cerebellum, deep nuclei, and/or cerebrum. This is not to say somatosensory information does not affect human movement complexity, since vibrating insoles operating at subsensory levels helped to restore center of pressure complexity in persons suffering from stroke, diabetic neuropathy, and the elderly [Costa et al., 2007; Priplata et al., 2006].

This application has expanded to running. At speeds above or below preferred, and also in novice runners, time-dependent variability was more predictable than speeds at preferred speed and in trained runners (Jordan et al., 2007; Nakayama et al., 2010). A more recent study found that fatigue and a history of previous injury both result in more random stride times as compared to healthy runners and an unfatigued state (Meardon et al., 2011), indicating the mover may have become less adaptable to possible environmental challenges. Cignetti et al. (2009) found a similar effect during cross-country skiing. Therefore, it seems that with supracortical diseases, injury, or fatigue, gait becomes more noisy, or random, but during less automated novel conditions, our system relies on more periodic, rigid control. Healthy, preferred, undiseased gait lies in the middle with optimal variability (Stergiou & Decker, 2011). When operating in that ideal, middle range, Stergiou & Decker (2011) assert that one can "...make flexible adaptations to everyday stresses placed on the human body."

Minimalist shoes and barefoot (BF) running are gaining popularity. A 2012 survey found 54% of runners reported switching footwear or foot strike style within the past 12 months,

mostly in hopes of decreasing injury risk (Goss & Gross, 2012). However, performing these non-preferred running patterns may render the runner less adaptable during their time of transition and susceptible to injury. As such, our purpose was to observe if stride complexity differs between habitual and novel running conditions: shod rearfoot strike (RFS), shod mid/forefoot strike (FFS), and barefoot (BF) running. We hypothesized stride time complexity will become more predictable when runners run with a novel foot strike pattern or when BF.

## Methods

Forty-two runners (21 each habitual rearfoot (hRF) and habitual mid/forefoot (hFF) strikers) participated in the study, which was approved by the university's Institutional Review Board. Written informed consent was obtained prior to participation. Because of missing data or difficulty in consistently auto-identifying impact acceleration peaks, thirty-three runners were included in the analysis. Subject characteristics are shown in Table 1. While there were more men in the hFF group, participants were not significantly different on the measured characteristics (all variables  $p \geq 0.15$ ).

**Table 1.** Subject characteristics between groups.

	<b>Sex (M, W)</b>	<b>Age (yrs)</b>	<b>Height (m)</b>	<b>Mass (kg)</b>	<b>Weekly Mileage (mi/wk)</b>	<b>Running Velocity (m/s)</b>
<b>hRF</b>	5,12	25±9	1.73±0.09	69.7 ±1.6	29±22	2.92±0.44
<b>hFF</b>	11,5	23±5	1.75±0.09	67.4±13.5	20±13	3.06±0.67

A tri-axial accelerometer (X250-2, Gulf Coast Data Concepts, LLC, Waveland, MS) was secured to the anteriomedial distal shank. Data were captured at 512 Hz and low-pass filtered at 70 Hz. Participants warmed-up on a treadmill for at least two minutes at their preferred speed. All shod conditions were performed in the runner's own shoes. Next, participants ran on the treadmill for 8 minutes each at their preferred speed for the following conditions: preferred foot strike (shod), non-preferred (i.e., converted) foot strike (shod), and barefoot (socks-only). They were not told which foot strike to use for the barefoot condition; if they asked, they were told to use "whatever comes natural." The researcher told them they could stop at any point if there was too much discomfort during the barefoot condition. The same preferred speed was used for all three conditions. Participants were visually observed to make sure they used the correct foot strike style during the FFS and RFS conditions.

A custom written Matlab program (8.4.0. 150421, R2014b, Natick, MA, USA) was used to auto identify impact peaks. If auto-identification did not successfully select all impact peaks or it selected incorrect points, we manually removed the incorrect points and added the correct points. Stride time was calculated based on time between impact peaks. Another Matlab program (fastdfa.m; Little et al., 2006) was used to calculate the scaling coefficient  $\alpha$  from detrended fluctuation analysis (DFA). If long range correlations exist in a signal,  $\alpha$  will be larger than 0.5. If the dependence of one value on the value of another prior point(s) decays according to the power law,  $\alpha$  will fall between 0.5 and 1.0. If long range correlations are present but there is no power law-type decay, then  $\alpha$  will be  $>1.0$ . Perfectly random data will yield  $\alpha \sim 0.5$ , and anti-correlated data (e.g., long, short, long, short, long stride) will give an  $\alpha < 0.5$  (Peng et al., 1995).

A repeated-measures ANOVA (2 (group)  $\times$  3 (condition)) was used to identify significant differences between groups and conditions for  $\alpha$ . Significance was evaluated at 0.05. If sphericity was violated, the Greenhouse-Geisser correction was used. Pairwise comparisons were assessed using a Sidak-adjustment. Cohen's  $d$  was calculated to compare differences between conditions using GPower 3.1. Differences of approximately 0.2, 0.5, and 0.8 were considered small, medium, and large effects, respectively (Cohen, 1992).

## Results

Over 600 strides were analyzed per condition (RF:  $614 \pm 51$ , FF:  $637 \pm 47$ , BF:  $630 \pm 84$ ). During the BF condition, 4 of the hFF participants did not complete the full 8 minutes due to pain on the plantar surface of their feet. Those participants only ran 4:30, 5:30, 5:35, and 7:45 minutes. This still resulted in over 350 strides analyzed and  $\alpha$  did not differ from others who ran the full 8 minutes, so their data were included.

Stride time decreased an average of 3.5% in the BF condition compared to runner's preferred foot strike. For hRF, stride time significantly shortened from RFS to FFS to BF, while for hFF, stride time during BF was significantly shorter than either shod condition (Table 2).

**Table 2.** Mean ( $\pm$ SD) stride time (s) for the groups during the three conditions.

	<b>RFS</b>	<b>FFS</b>	<b>BF</b>
<b>hRF</b>	$0.722 \pm 0.040$	$0.713 \pm 0.039^*$	$0.694 \pm 0.040^{*^{\wedge}}$
<b>hFF</b>	$0.729 \pm 0.044$	$0.733 \pm 0.049$	$0.706 \pm 0.048^{*^{\wedge}}$

\*Significantly different from RFS ( $p < 0.05$ ).

$^{\wedge}$  Significantly different from FFS ( $p < 0.05$ ).

Peak impact accelerations significantly increased during the novel conditions for hRF (Table 3) but they only significantly increased from RFS to BF for the hFF.

**Table 3.** Mean ( $\pm$ SD) peak impacts (g's) for the groups during the three conditions.

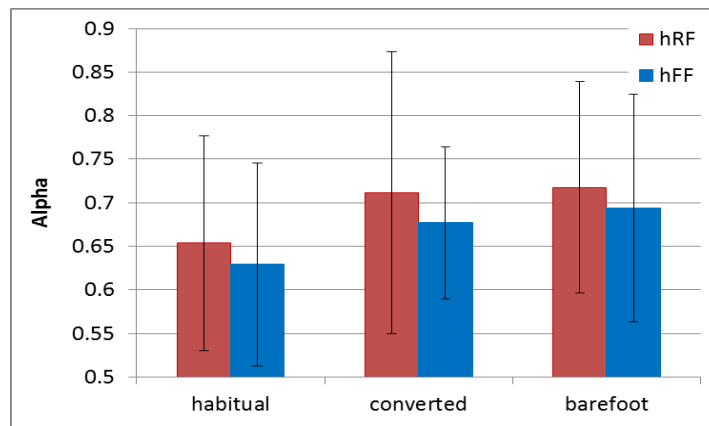
	<b>RFS</b>	<b>FFS</b>	<b>BF</b>
<b>hRF</b>	4.45 $\pm$ 1.95	6.27 $\pm$ 2.60*	6.66 $\pm$ 3.37*
<b>hFF</b>	5.34 $\pm$ 2.84	6.25 $\pm$ 2.27	6.98 $\pm$ 3.26^

\*Significantly different from RFS ( $p < 0.05$ )

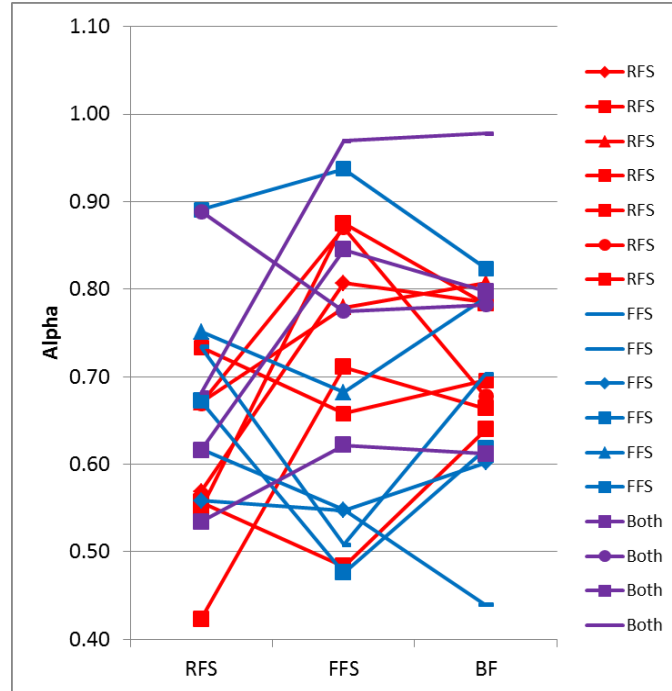
^ Significantly different from FFS ( $p < 0.05$ )

During the BF condition for hRF, 7 used a RFS the entire time, 6 used a FFS the entire time, and 4 switched between foot strike styles (Figure 2). In the hFF group, one participant switched between foot strikes styles while all the rest used a FFS the entire time.

There was not a significant group  $\times$  condition interaction ( $p = 0.969$ ) or group main effect ( $p = 0.410$ ) for  $\alpha$ . Condition bordered on significance ( $p = 0.080$ ). Despite insignificance, the effect sizes between the habitual and novel conditions displayed medium effect sizes (Table 4). Alpha was larger during the novel conditions compared to preferred foot strike for both groups.



**Figure 1.** Alpha for the two groups during the three conditions.



**Figure 2.** Alpha for all 17 hRF. The colors distinguish which foot strike(s) runners used during the BF condition. Red=RFS, Blue=FFS, and Purple=Both.

**Table 4.** Mean ( $\pm$ SD) alpha for the groups during the three conditions, and effect sizes (d).

	Alpha			Effect Size (d)		
	Habitual	Converted	Barefoot	H v. C	H v. BF	C v. BF
<b>hRF</b>	0.65 $\pm$ 0.12	0.71 $\pm$ 0.16	0.72 $\pm$ 0.12	0.47	0.50	0.04
<b>hFF</b>	0.63 $\pm$ 0.12	0.68 $\pm$ 0.09	0.69 $\pm$ 0.13	0.41	0.55	0.13

## Discussion

We aimed to see if the long range correlation of stride times change during novel running conditions. Our data support our hypothesis— $\alpha$  moderately increased when hRF and hFF ran with the opposite foot strike (shod) and BF as compared to their preferred shod foot strike style. This corroborates other findings that performing novel conditions (running with little experience) or running at non-preferred speeds leads to stronger long range correlations (Jordan et al., 2007; Nakayama et al., 2010). Our results may indicate runners moved from a steady state of optimal movement adaptability, with greater dynamical degrees of freedom during their preferred foot strike pattern, to being slightly more rigid while running with the non-preferred conditions. This may manifest as increase muscular co-contraction, which has been observed early in motor learning (Osu et al., 2002). If runners are less adaptable during these novel conditions, they may

be at greater risk of injury. Thus, runners may benefit from minimizing exposure to dynamic, unstable environments (e.g., trail running or amongst people or obstacles) that require more adaptation, and transition to new footwear or foot strike styles on a treadmill. Or the runners may benefit from additional practice of the novel conditions. There, however, have not been any prospective studies documenting if change in  $\alpha$  during non-preferred movement conditions leads to injury, so this is speculative. However, several authors have noted that during a fatigued state, which may be considered non-preferred, that risk of injury increases ((Donahue and Sharkey, 1999; Yoshikawa et al., 1994; Arndt et al., 2002; Milgrom et al., 2007; Clansey et al., 2012)), so this hypothesis is not unreasonable.

The difference in  $\alpha$  from preferred was approximately 0.05, which may not be large enough to negatively affect a runner's adaptability. For instance, the difference between novice and experienced runners, slow and preferred speeds, and injured and healthy controls was 0.13-0.17 (Jordan et al., 2006; Nakayama et al., 2010; Meardon et al., 2011). An acceptable adaptability threshold has not been established, so it is possible our difference between conditions was not clinically meaningful. Our average  $\alpha$  values for the preferred condition ( $\sim 0.64$ ) were also lower than previously reported for treadmill running (Jordan et al., 2006; 2007; Nakayama et al., 2010) ( $\sim 0.73$ - $0.78$ ). This may partially be due to our inclusion of impact peaks that were not automatically-identified (i.e.,  $\alpha$  from auto-identification was usually higher than  $\alpha$  when additional manually-identified impact peaks (and stride times) were included). Typically the added peaks were significantly shorter stride times.

The overall increase in  $\alpha$  for the novel conditions was not universal; five hRF and two hFF had lower values for both novel conditions, and several others had lower values for at least one novel condition. It is likely, therefore, that multiple factors influence stride time predictability, possibly increasing, decreasing, or maintaining  $\alpha$ . As mentioned previously, fatigue decreases  $\alpha$  (Meardon et al, 2011), which is a viable explanation considering most runners experience some plantarflexor fatigue or pain when switching to a FFS or BF (Hryvniak et al., 2014). Fatigue could have mitigated some of the effects of the novel condition, which is expected to increase  $\alpha$ . Additionally, pain (which might be an extreme form of fatigue) was starting to occur in runners who did not complete the 8 minutes of BF running (and is suspected to have occurred in most others who switched to a FFS when BF) may have affected  $\alpha$ . We, however, did not objectively measure fatigue or pain so these are speculations. Familiarity would

help keep  $\alpha$  constant, but it alone did not explain the differential response for all runners, either. Twelve runners reported at least some BF training in high school, but  $\alpha$  was not necessarily more similar in BF compared to their preferred shod running. Five hFFs used to be rearfoot strikers but converted to a FFS. This did not seem to result in  $\alpha$  being more similar for RFS in that subset of hFF runners. In fact,  $\alpha$  increased minimally or moderately for four of those five runners during RFS. Most hRF runners likely had some experience with FFS running when sprinting, which may partly explain a smaller average change in  $\alpha$ . Nearly all runners changed their stride time (thus stride length) to varying degrees during the novel conditions, which may have also influenced  $\alpha$ . We thought, perhaps,  $\alpha$  during BF would be more similar to the shod condition in which foot strike was similar, but that does not appear to be the case (see Figure 2). Hence, the multiple factors affecting long range correlations seem to make the results slightly ambiguous.

As others have surmised, looking at the time-dependent variability provides more information than looking at central variability parameters, such as coefficient of variability (CV) (Stergiou & Decker, 2011). In our sample, CV was similar ( $p>0.05$ ) between conditions for hRF (2.5, 2.6, and 2.5%, respectively, for habitual, converted, and BF) and decreased ( $p<0.05$ ) for the novel conditions for hFF (3.6, 3.1, and 3.1%, respectively). One would come to different conclusions for hRF whether learning has occurred and one's movement pattern is stable whether using CV or  $\alpha$  to measure variability. Therefore, it is important to also consider the time-dependent repeatability.

There are a few limitations to our study. We only used visual (and auditory) observation to confirm the correct foot strike style was being used, which is less accurate than using motion capture systems (Altman & Davis, 2012). Additionally, for runners whose preferred shod foot strike was not easily identifiable, we used a high-speed video camera to distinguish foot strike style, so we are fairly confident in our foot strike classification. We used accelerometer data to calculate stride time. Because some impact peaks were not easily identified, some stride times were omitted from analysis, which may have affected  $\alpha$ .

## **Conclusion**

Altering foot strike and/or running BF affects the underlying variability of stride time, resulting in stride interval timing that becomes more patterned. This may make the runner less adaptable, possibly increasing their risk of injury, but there was still a large amount of individual variability in response to the novel conditions.

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## CHAPTER 7

### TIBIAL STRESSES DURING BAREFOOT AND SHOD RUNNING WITH NORMAL AND SHORTENED STRIDE LENGTHS

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#### Abstract

Forces at the ankle appear to be larger with a mid/forefoot strike (FFS) compared to a rearfoot strike (RFS) when running, but they decrease at the knee. We do not know how this affects stresses in the distal tibial, which is a common location of stress fractures. Our purpose was to estimate subject-specific stresses in the distal tibia in habitual rearfoot (hRF) and habitual mid/forefoot (hFF) runners when using their preferred and non-preferred foot strike styles when shod and during barefoot (BF) running with normal and shorter stride lengths (SL). Seventeen runners performed overground running trials at 3.5 m/s for the following 6 conditions: 1) RFS at preferred SL (PSL), 2) FFS at PSL, 3) BF at PSL, 4) RFS at -8%PSL, 5) FFS at -8%PSL, and 6) BF at -8% PSL. Information from a CT scan of runners' distal tibia was combined with kinematic and kinetic data to estimate bone stresses. Peak compressive, tensile, and shear stresses were larger during shod FFS than shod RFS or BF. Typically stresses were smallest for RFS. Only shear stress significantly decreased with a shorter SL. Unless one's bones have adapted to the greater loads of FFS and BF, it may lead to injury. Since an 8% shorter SL did not sufficiently decrease normal stresses, a more drastic change (e.g., 10%) may be required to decrease injury risk.

#### Introduction

Few studies have compared peak bone stresses or strains during shod rearfoot and mid/forefoot strike running or shod versus barefoot. With the surge of interest in minimalist shoes/barefoot running (Rothschild, 2012; Hryvniak et al., 2014; Goss & Gross, 2012), runners

need to know how a running style transition will affect their risk of injury, primarily stress fractures.

Bones stresses or strains provide a better estimate of lower extremity loading than ground reaction forces, joint moments, strains from gauges, and EMG, which are the primary variables reported in the literature for the abovementioned running styles. For instance, ground reaction forces can only indicate the net acceleration of the body during ground contact. If only considering vertical and posterior ground reaction forces, Ekenman et al. (1998) would have concluded that landing on the forefoot (FF) during a 30-cm forward jump results in lower loading on the body as compared to landing on one's heel. Interestingly, though, peak tibial strains were higher during the FF landing, especially at the posteromedial distal third location. This most likely was a result of greater contraction of the triceps surae muscles. Strain gauges have their limitations, too, namely that the values are highly location-specific and they are difficult to place on locations that are covered by muscle. Meardon & Derrick (2014) purport that peak compressive (C) and tensile (T) stresses occur on the posterior and anterior tibial cortex, respectively, during running. At a more medial location, closer to where strain gauges are typically mounted, they estimated peak strains of approximately  $\pm 10$  MPa, which closely agrees with in vivo data (Burr et al., 1996; Ekenman et al., 1998) but vastly underestimated the actual peak stress values. Joint moments as calculated from inverse dynamics are limited in that they only indicate the net moment (i.e., the sum of two opposing moments), so co-contraction is not accounted for when this is used to quantify joint loading. For example, a positive extensor moment at the knee would be interpreted as the knee extensors being active while the antagonistic knee flexors are inactive. However, use of EMG would demonstrate this is not true during midstance of running (Chumanov et al., 2012; Giandolini et al., 2013; Shih et al., 2013; Teng & Powers, 2014), when there is simultaneous activation of the knee extensors (quadriceps) and knee flexors (hamstrings and gastrocnemius). Finally, relying on EMG to quantify loading is unreliable as well, since it only picks up the electrical signal across a muscle fiber. It does not take into consideration whether the muscle is lengthening or contracting, its current fiber length, or the velocity at which it is contracting—all variables that affect how much force that muscle can produce. Additionally, surface EMG can only measure muscle activity of superficial muscles. Therefore, using musculoskeletal modeling to calculate joint contact forces, which is

the vector sum of the joint reaction forces and muscle forces, provides estimates of loads the joints are experiencing due to gravity, segment accelerations, and muscle forces.

Only two preliminary studies have estimated tibial stresses or strains during different running styles. Derrick Edwards & Rooney (2012) found greater peak stresses in the distal tibia cross section in 3 out of the 4 quadrants during shod midfoot/forefoot strike (FFS) versus shod rearfoot strike (RFS). The limitations of their results include a generic ellipsoid model used for the tibia and the running speeds between groups not being identical. Altman & Davis (2012) improved upon this and incorporated information from CT scans of the tibia while runners ran at the same speed (3.5 m/s). In their five subjects, they found small trends towards greater peak normal strains in BF running, while the largest peak strain rates occurred in shod FFS. Shod RF was lowest in both instances. Unfortunately, they only had habitual rearfoot strikers run with both foot strike styles. Since bone adapts to the dynamic loads placed on it (Lanyon & Rubin, 1984; Bhatia et al., 2015; Troy et al., 2013), rearfoot strikers' lower extremity bones may be different than mid/forefoot strikers' bones, thus resulting in different stresses and strains.

Our purpose was to address the limitations of the prior two preliminary studies by comparing distal tibia bone stresses in habitual rearfoot strikers (hRF) and habitual mid/forefoot strikers (hFF) during running with a 1) shod RFS, 2) shod FFS, and 3) barefoot using a FFS (BF). The secondary purpose was to see if these stresses decrease with an 8% shorter stride length (SL), as we wanted to see if shortening one's SL could decrease loads to the same extent as switching foot strike styles. Based on results from Altman & Davis (2012) and Lanyon et al. (1975) who found higher strains during BF than shod running, we hypothesized that peak compressive, tensile, and shear stresses would be greatest during BF running and lowest in shod RFS. We hypothesized the stresses would decrease when using a shorter SL.

## **Methods**

### **Participants**

Male and female recreational or competitive runners age 18-40 were recruited. Additional inclusion criteria included: 1) running 10+ miles per week over the last 3+ years, 2) injury-free within the last three months, 3) have not switched foot strike styles within the past two years, 4) did not have lower-extremity surgery that would affect their running, 5) not a tobacco user, and 6) have not suffered from amenorrhea. hRF and hFF runners were recruited, matching on sex, age, mass, and weekly running mileage.

### Protocol

The study was approved by the university's Institutional Review Board, and permission to conduct the CT scans was granted from the Iowa Department of Public Health. Written informed consent was obtained prior to participation. Midfoot and forefoot strikers were grouped together as hFF because of the scarcity of forefoot strikers (Cavanagh & LaFortune, 1980). Twenty-eight individual reflective markers were placed on the lower extremity and shoes (sacrum and C7, and bilaterally on the acromion, ASIS, greater trochanter, lateral thigh, anterior distal thigh, lateral femoral condyle, anterior proximal & distal leg, posterior leg, lateral malleolus, dorsifoot, lateral foot, and heel). Four more markers were placed on the medial malleoli and femoral condyles for the standing calibration trial used to identify knee and ankle joint centers.

Participants performed a brief warm up on a treadmill. Then they ran at 3.5 m/s for two minutes each with a shod RFS, shod FFS, and BF. At the end of the 2 minutes, 20 timed strides used to calculate preferred stride length (PSL) for each condition. Participants wore their own running shoes (none wore barefoot-simulating shoes).

Participants then completed seven overground running trials down a 30 m runway, all at 3.5 m/s ( $\pm 3\%$ ) for 6 conditions: 1) shod RFS at PSL, 2) shod FFS at PSL, 3) BF at PSL, 4) shod RFS at  $-8\%$ PSL, 5) shod FFS at  $-8\%$ PSL, and 6) BF at  $-8\%$ PSL. The PSL used was that of their habitual foot strike style. This was done to determine the independent effect of shoes and foot strike style. Overground step length was marked with athletic tape. Participants always performed their preferred shod foot strike conditions first, followed by the opposite shod foot strike, followed by BF. The order of presentation of the SL conditions was counterbalanced. Shortened SL trials were considered acceptable if SL was within  $\pm 4\%$  of the target. Strike index (SI) was calculated as the average center of pressure location during the first 10 ms of stance and reported as a percentage of foot length from the posterior calcaneus (Cavanagh & LaFortune, 1980). A foot strike with an SI  $< 33.3\%$  was a RFS, and FFS had a SI  $> 33.3\%$ . The only trials included in the analysis were ones with the correct foot strike style based on SI, within the tolerated speed and step length, and no visual targeting. At the conclusion of the study, all runners completed a detailed questionnaire about their running and health habits.

### Equipment

Kinematic data were collected at 200 Hz with an 8-camera Vicon system (Vicon MX, Vicon, Centennial, CO, USA) and low-pass filtered at 12 Hz. Cardan angles were calculated for

the lower extremity joints using a flexion/extension, ab/adduction, internal/external rotation order. Kinetic data were collected at 1000 Hz by an in-ground AMTI force platform (AMTI, Watertown, MA) and low-pass filtered at 20 Hz. SL was calculated as the anterior-posterior distance between the left heel markers.

#### Data Processing and Musculoskeletal and Finite Element Models

Standard inverse dynamics was used to calculate joint reaction forces and moments, which are presented in the distal segment coordinate system. Estimates of segment mass, center of mass location, and moments of inertia were obtained using participants' anthropometrics (Vaughan et al., 1992). Leg length (LL) was defined as the distance from the superior border of the greater trochanter to the floor. Lower extremity muscle properties were obtained using a musculoskeletal model (Arnold et al., 2010), implemented within custom Matlab software (R2014b, Natick, MA, USA). Muscle forces were estimated using static optimization and constrained using force-length and force-velocity adjusted maximal values. Because there was insufficient muscle force available to match the peak moments from inverse dynamics, maximum allowable muscle force was increased by a factor of 1.75. A set of muscle forces that matched moments derived from inverse dynamics and that minimized the sum of muscle stresses squared was selected (Glitsch & Baumann, 1997). Sagittal plane moments at the hip, knee, and ankle, and frontal plane moments at the hip and ankle were used for the optimization. Muscle forces were summed with joint reaction forces to estimate joint contact forces.

A 2-mm thickness CT scan (Toshiba Prime 40) was obtained for each participant's right leg using the following settings: 120 kV, 50 mAs 100 mA with 0.5 sec/rotation. A bone standard filter in the CT software was used. The scan was taken one-third of the way from the ankle to the knee, because it is a common stress fracture site (Crossley et al 1999). Based on the relationship between the Hounsfield scale and apparent bone density, the elastic modulus (E) of all elements was determined from the CT image. Forces and moments acting at the level of the CT scan were estimated and used as inputs to a finite element mesh to estimate bone stresses (VA-BATTS, Version 3; Kourtis et al, 2008). A 600-element mesh (60 perimeter x 10 deep) was fit to the cross-section after a grayscale auto-identification of bone was used to identify the outer and inner perimeters.

#### Statistical Analysis

Peak compressive (C), tensile (T), and shear (S) stresses during stance phase were our variables of interest. Values were averaged across acceptable trials for each condition. A 2×2×3

repeated-measures MANOVA (habitual group  $\times$  foot strike  $\times$  SL) was performed in SPSS Statistics, Version 21.0 (IBM SPSS Statistics for Windows, Armonk, NY). Significance was set to  $\alpha=0.05$ . If the MANOVA was significant, the univariate ANOVAs were subsequently evaluated at a 0.05 level. If sphericity was violated, the Greenhouse-Geisser correction was used. Pairwise comparisons were made using a Sidak adjustment. To facilitate comparison of meaningful differences despite the small sample size, effect size ( $d$ ) was also calculated using GPower 3.1. Differences of approximately 0.2, 0.5, and 0.8 were considered small, medium, and large effects, respectively (Cohen, 1992).

## Results

Nine hRF and 9 hFF were recruited. All subjects completed the running analysis, but one hFF has not completed his CT scan at this point. Groups did not significantly differ based on age, height, mass, BMI, or weekly mileage (all  $p>0.417$ ). Comparisons by sex between groups also were non-significant (all  $p>0.200$ ) (Table 1). Bone geometry and properties are given in Table 2.

**Table 1:** Subject characteristics (mean  $\pm$ SD).

	<i>hRF</i>			<i>hFF</i>		
	<i>Male</i>	<i>Female</i>	<i>Overall</i>	<i>Male</i>	<i>Female</i>	<i>Overall</i>
<b>Participants</b>	4	4	9	5	4	9
<b>Age (yr)</b>	23 $\pm$ 5	25 $\pm$ 5	24 $\pm$ 5	28 $\pm$ 7	24 $\pm$ 6	26 $\pm$ 7
<b>Height (m)</b>	1.82 $\pm$ 0.06	1.64 $\pm$ 0.08	1.74 $\pm$ 0.11	1.78 $\pm$ 0.07	1.64 $\pm$ 0.07	1.71 $\pm$ 0.10
<b>Leg length (m)</b>	0.969 $\pm$ 0.027	0.859 $\pm$ 0.043	0.920 $\pm$ 0.066	0.953 $\pm$ 0.046	0.887 $\pm$ 0.025	0.924 $\pm$ 0.050
<b>Mass (kg)</b>	68.9 $\pm$ 5.8	56.6 $\pm$ 11.3	63.4 $\pm$ 10.3	67.9 $\pm$ 12.8	58.6 $\pm$ 10.6	63.8 $\pm$ 12.2
<b>BMI</b>	20.7 $\pm$ 1.1	20.8 $\pm$ 2.1	20.8 $\pm$ 1.5	21.4 $\pm$ 2.6	21.7 $\pm$ 2.1	21.5 $\pm$ 2.3
<b>Weekly mileage (mi/wk)</b>	45 $\pm$ 32	18 $\pm$ 11	33 $\pm$ 27	41 $\pm$ 43	28 $\pm$ 16	35 $\pm$ 32



**Table 2:** Bone characteristics of the runners (mean  $\pm$ SD).

	<i>hRF</i>			<i>hFF</i>		
	<i>Male</i>	<i>Female</i>	<i>Overall</i>	<i>Male</i>	<i>Female</i>	<i>Overall</i>
<b>Average E (GPa)</b>	20.3 $\pm$ 1.3	22.0 $\pm$ 0.3	21.1 $\pm$ 1.3	20.9 $\pm$ 1.1	21.4 $\pm$ 0.4	21.1 $\pm$ 0.9
<b>CSA (mm<sup>2</sup>)</b>	364 $\pm$ 46	275 $\pm$ 65	319 $\pm$ 70	329 $\pm$ 42	288 $\pm$ 51	311 $\pm$ 49
<b>I<sub>ML</sub> (mm<sup>4</sup>)</b>	15,799 $\pm$ 2654	9780 $\pm$ 4085	12,789 $\pm$ 4530	13,422 $\pm$ 3358	9807 $\pm$ 3007	11,815 $\pm$ 3558
<b>I<sub>AP</sub> (mm<sup>4</sup>)</b>	20,147 $\pm$ 3394	12,288 $\pm$ 5515	16,218 $\pm$ 5968	16,348 $\pm$ 3,268	12,944 $\pm$ 5928	14,835 $\pm$ 4662
<b>Outer ML diameter (mm)</b>	26.8 $\pm$ 0.7	22.7 $\pm$ 1.9	24.7 $\pm$ 2.6	25.6 $\pm$ 1.4	23.8 $\pm$ 3.5	24.8 $\pm$ 2.5
<b>Outer AP diameter (mm)</b>	24.2 $\pm$ 0.3	21.4 $\pm$ 1.7	22.8 $\pm$ 1.9	23.5 $\pm$ 1.8	21.0 $\pm$ 1.9	22.4 $\pm$ 2.2
<b>Cortical ML thickness (mm)</b>	7.8 $\pm$ 0.5	6.3 $\pm$ 1.7	6.3 $\pm$ 2.7	7.6 $\pm$ 0.9	7.2 $\pm$ 1.2	7.5 $\pm$ 1.0
<b>Cortical AP thickness (mm)</b>	6.2 $\pm$ 0.8	5.3 $\pm$ 1.0	5.1 $\pm$ 2.1	6.0 $\pm$ 0.8	5.3 $\pm$ 0.6	5.7 $\pm$ 0.8

E: elastic modulus, CSA: cross sectional area, I<sub>ML</sub>: mediolateral area moment of inertia, I<sub>AP</sub>: anteroposterior area moment of inertia, ML: mediolateral, AP: anteroposterior

### Preferred Stride Lengths

The two groups of runners had similar PSL for their respective habitual foot strike during treadmill running (hRF:  $269 \pm 16\%$  LL, hFF:  $262 \pm 14\%$  LL,  $p=0.347$ ) (Table 3). All runners shortened their SL when running BF as compared to their habitual foot strike (avg:  $0.11 \pm 0.05$  m,  $p<0.001$ ; hRF:  $5.4 \pm 1.9\%$  PSL, hFF:  $3.3 \pm 1.7\%$  PSL). hRF used significantly shorter strides from RFS to FFS to BF, while hFF chose similar SLs for RFS and FFS but a shorter SL when BF.

**Table 3:** Preferred stride length calculated during treadmill running for the 3 conditions, given in m and as a % of leg length (LL) (mean  $\pm$ SD). Runners used their habitual stride length in all experimental conditions.

	hRF			hFF			Overall		
	<i>RF</i>	<i>FF</i>	<i>BF</i>	<i>RF</i>	<i>FF</i>	<i>BF</i>	<i>RF</i>	<i>FF</i>	<i>BF</i>
<b>Stride length (m)</b>	2.470* $\pm 0.189$	2.428*^ $\pm 0.187$	2.336 $\pm 0.173$	2.398* $\pm 0.127$	2.417* $\pm 0.126$	2.337 $\pm 0.118$	2.434* $\pm 0.161$	2.423* $\pm 0.155$	2.337 $\pm 0.144$
<b>Stride length (%LL)</b>	269* $\pm 16$	264*^ $\pm 14$	254 $\pm 14$	260* $\pm 16$	262* $\pm 14$	254 $\pm 16$	264* $\pm 16$	263* $\pm 13$	254 $\pm 15$

\* Significantly different compared to BF ( $p<0.05$ )

^ Significantly different compared to RF ( $p<0.05$ )

### MANOVA Results

The three-way interaction was not significant ( $p=0.907$ ), nor was group  $\times$  SL ( $p=0.286$ ), SL  $\times$  run style ( $0.097$ ), or group  $\times$  run style ( $p=0.765$ ). There was not a main effect for group ( $p=0.712$ ). SL ( $p=0.010$ ) and run style ( $p<0.001$ ) were significant.

Example stress curves during stance are shown in Figure 2. All variables were significant for run style (peak T  $p=0.022$ , peak C  $p=0.003$ , peak S  $p=0.029$ ). Pairwise comparisons revealed that peak T, C, and S stresses were larger for shod FFS than shod RFS (all  $p \leq 0.015$ ) (Tables 4 & 5). In general, there were small to large effect sizes for peak stresses being larger in BF vs. RFS, with the exception of a medium decrease in T stresses when hRF switched to BF (Table 5). BF generally had smaller peak stresses than FFS, with more pronounced differences for hRF than hFF.

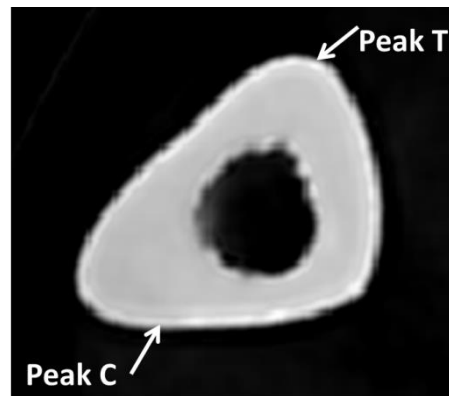
Peak S significantly decreased with an 8% shorter SL ( $0.003$ ), peak T bordered on significance ( $p=0.090$ ,  $d=0.26$ ), and compressive stresses did not change ( $p=0.394$ ).

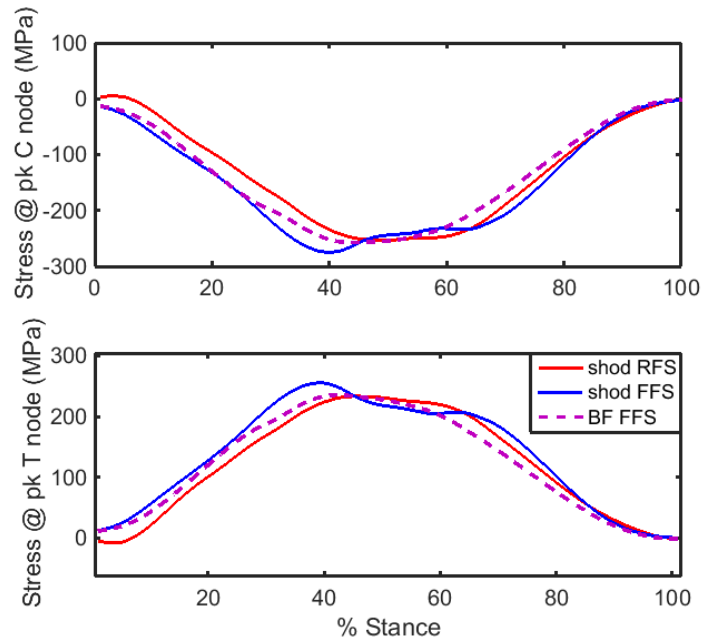
**Table 4.** Mean  $\pm$ SD peak tensile (T), compressive (C), and shear (S) stresses for the 2 groups and 6 conditions.

	hRF	hFF	hRF	hFF	hRF	hFF
	<i>Peak T</i>		<i>Peak C</i>		<i>Peak S</i>	
<b>RF</b>	193 $\pm$ 42	215 $\pm$ 40	-189 $\pm$ 38	-211 $\pm$ 40	9 $\pm$ 2	9 $\pm$ 3
<b>RF -8%</b>	194 $\pm$ 44	211 $\pm$ 39	-192 $\pm$ 37	-207 $\pm$ 38	9 $\pm$ 3	9 $\pm$ 2
<b>FF</b>	204 $\pm$ 48	228 $\pm$ 51	-207 $\pm$ 46	-231 $\pm$ 45	12 $\pm$ 4	11 $\pm$ 3
<b>FF -8%</b>	205 $\pm$ 52	226 $\pm$ 46	-210 $\pm$ 50	-230 $\pm$ 41	11 $\pm$ 4	10 $\pm$ 2
<b>BF</b>	185 $\pm$ 35	222 $\pm$ 56	-193 $\pm$ 34	-228 $\pm$ 49	12 $\pm$ 3	11 $\pm$ 3
<b>BF -8%</b>	183 $\pm$ 44	212 $\pm$ 56	-193 $\pm$ 40	-220 $\pm$ 43	10 $\pm$ 3	10 $\pm$ 3

**Table 5.** Effect size (d) between PSL conditions.

	hRF			hFF		
	<i>RF v. FF</i>	<i>RF v. BF</i>	<i>FF v. BF</i>	<i>RF v. FF</i>	<i>RF v. BF</i>	<i>FF v. BF</i>
<b>Peak T</b>	1.26 <sup>F</sup>	0.48 <sup>R</sup>	0.88 <sup>F</sup>	0.83 <sup>F</sup>	0.28 <sup>B</sup>	0.23 <sup>F</sup>
<b>Peak C</b>	1.75 <sup>F</sup>	0.29 <sup>B</sup>	0.70 <sup>F</sup>	1.57 <sup>F</sup>	0.90 <sup>B</sup>	0.12 <sup>F</sup>
<b>Peak S</b>	0.96 <sup>F</sup>	0.84 <sup>B</sup>	0	0.84 <sup>F</sup>	0.48 <sup>B</sup>	0

<sup>F</sup> Values larger for FFS<sup>R</sup> Values larger for RFS<sup>B</sup> Values larger for BF**Figure 1.** Example tibia cross-section with approximate locations of where peak T and C stresses occur during running.



**Figure 2.** Representative normal stresses during stance for the 3 PSL conditions for a hRF at the two nodes where peak tensile (T) and compression (C) stresses occurred, which was  $68^\circ$  and  $232^\circ$  relative to a right horizontal, respectively. Positive values are T and negative values are C.

## Discussion

The purpose of this study was to compare peak stresses in the distal tibia during shod RFS, shod FFS, and BF running. Our hypotheses were only partially supported—peak stresses were smallest for shod RFS for all variables as hypothesized, except peak T stresses for hRF were lowest for BF. All stresses were largest for shod FFS, not BF, which does not support our hypothesis. Our data for peak T and C stresses do not support the trends previously observed in 5 hRF (Altman & Davis, 2012), which may partially be due to small sample sizes. With a shorter SL, only S stress significantly decreased, which only partially supports our hypothesis.

The greater work performed by the plantarflexors during FFS and BF (Williams et al., 2012; Paquette et al 2013) likely resulted in the greater tibial stresses since the gastrocnemius and soleus cross the distal third of the tibia. Altman and Davis (2012) hypothesized perhaps their shod FFS strain rates were larger than BF because a larger plantarflexion angle is needed to accommodate the shoe midsole. Our data does not support this, however, as the foot was approximately  $4^\circ$  more plantarflexed at contact when BF (FFS:  $-4 \pm 4^\circ$  v. BF:  $-8 \pm 5^\circ$ ).

Interestingly, peak plantarflexion moment was greatest for the BF PSL condition (RFS:  $0.273 \pm 0.027$ , FFS:  $0.316 \pm 0.031$ , BF:  $0.324 \pm 0.036$  BW\*m), but that did not equate to the largest stresses. Perhaps the larger vertical reaction forces for FFS versus BF at the knee ( $2.05 \pm 0.17$  v.  $2.00 \pm 0.16$  BW,  $p=0.017$ ) and ankle ( $2.61 \pm 0.15$  v.  $2.56 \pm 0.17$  BW,  $p=0.005$ ) partially explain the differences. The reaction forces also tended to be larger for FFS than RFS (knee:  $2.01 \pm 0.16$  BW,  $p=0.098$ ; ankle:  $2.51 \pm 0.16$  BW,  $p=0.003$ ). These differences persisted despite all conditions utilizing the same SL, with only minor differences in mean SL (hRF: RFS condition was 1.3 and 2.3 cm longer than FFS and BF, respectively. hFF: FFS was 0.3 and 1.1 cm longer than RFS and BF, respectively). Taken together, it may not be advisable for shod runners to switch to a FFS in order to reduce tibial stresses, if the transition is done hastily. However, hRF may decrease T stress if they ran BF. A transition period of longer than 10 weeks is recommended to help avoid injury of the foot (Ridge et al., 2013). If transitioning to BF, it is important to use a mid- or forefoot strike since vertical ground reaction force loading rates can be extremely high and potentially injurious if using a RFS when BF (Cheung & Rainbow, 2014; DeWit et al, 2000; Stacoff et al., 2000; Lieberman et al. 2010). Alternatively, osteogenic stimulation is greater under higher loads and strain rates (Qin et al., 1998). So if done carefully, transitioning to a shod FFS may improve one's bone strength, although our small sample size does not indicate a trend towards hFF having stronger (higher elastic modulus) bones.

Further exploration into why peak T stress was lower for BF than RFS in hRF, when the opposite was true for hFF, revealed that peak vertical knee reaction force may account for the differences. It was lower when BF compared to RFS for hRF (1.96 versus 2.02 BWs) but was greater for BF in hFF (BF: 2.04 versus RFS: 1.99 BWs). Therefore, switching to BF using a FFS may be an alternative for hRF, especially if coupled with a naturally shorter SL. However, BF did not decrease C stresses and slightly increased S stresses, so it is not unanimously better.

The highest C stress consistently occurred in the posteromedial quadrant, which agrees with the location where stress fractures most commonly occur in runners (Crossley et al 1999; Nattiv et al 2013). Tensile stresses were highest in the anterolateral quadrant, which is another common location for stress fractures in runners (Nattiv et al., 2013). Our peak stresses as a proportion of ultimate stress would suggest that T stress fractures in the

anterolateral quadrant would be more common. This may indicate a limitation of our model estimations.

It has been suggested that decreased loads during BF running may be due to a self-selected shorter SL (Thompson et al., 2014). Since we controlled for SL, the differences observed in tibial stresses seem to be due to foot strike/footwear. It may not be appropriate, however, to parse out the effect of stride length, since runners naturally choose a 2.5-6.1% shorter SL when BF (De Wit et al., 2000; Bonacci et al 2013; Kerrigan et al., 2009; Warne & Warrington, 2013). During treadmill running, all of our 18 runners chose a shorter SL when BF (avg. decrease 4.3% relative to PSL for their habitual foot strike), so “true” BF running would presumably decrease loads even more, but not to the extent of the -8%PSL BF condition.

While these decreases in peak S stress during the shorter SL condition seems small (1 MPa), because of the exponential decrease in cycles to failure in bone as stress magnitude increases (Edwards & Derrick, 2011), use of a shorter SL may significantly decrease bony injury risk. The authors stated that a 10% change in stress resulted in a 75-100% change in cycles to failure, but this seems high. While some participants had peak normal stresses change by ~10% between conditions, on average the -8%PSL condition decreased T and C stresses by 1-4%, while switching from shod FF to either BF or shod RF resulted in 4-9% decreases. Edwards et al. (2009) noted that peak resultant tibial contact forces decreased 0.7 BW (5.2%) when using a 10% shorter stride length, which subsequently resulted in a 3-6% reduction in probability of tibial stress fracture when running 3 to 7 miles per day, respectively. Surprisingly, our peak axial ankle contact forces did not change relative to PSL, and with such a small change in peak stresses, an 8% shorter SL may not sufficiently decrease probability of tibial stress fracture. A larger 10% decrease may be required to be impactful (Edwards et al., 2009). Change in foot strike or footwear, however, seems to have a larger effect.

In general, our peak tibial stresses are high compared to the literature. As such, the relative differences are more important than the absolute stress values. Altman & Davis (2012) found peak stresses of approximately 100 MPa and -200 MPa (assuming an elastic modulus of 20 GPa) at midshaft. Our values are approximately twice as high for tensile stresses but similar for compressive stresses. Our peak ankle contact forces were

approximately similar (~6500 N) to theirs, suggesting our optimization routines used to estimate muscle forces was similar. The discrepancy may be how we transformed forces and moments acting at the level of the scan. Meardon and Derrick (2014) calculated peak stresses of 55 and -90 MPa when running at 4 m/s, which are lower than our values for two main reasons. The ellipsoid representation of the tibia they used underestimates stresses at the distal tibia, especially tensile stresses (Derrick et al 2015, unpublished data), and they only analyzed stresses at 4 points (anterior, posterior, medial, lateral), whereas we identified peak stresses from the entire cross-section. It is also unclear how similar our muscle optimization routines were, which could be another source of discrepancy.

Our stress estimates are also higher than in vivo strain gauge estimates. Burr et al. (1996) observed peak strains on the medial midshaft of approximately 13 MPa and -19 MPa when running at 3.82 m/s. Ekenman et al (1998) observed similar stresses (approx. 11 and -22 MPa) at the posteromedial distal tibia during running at an unreported self-selected speed. There are several plausible reasons our stresses are higher than in vivo: 1) we neglected the fibula's role in weight-bearing, 2) static optimization tends to overestimate muscle forces (Prilutsky et al., 1997), which will increase bending stresses, and 3) in vivo strain gauges are not typically placed in locations where stresses are largest (Derrick et al 2015 unpublished data). It may be disconcerting that our peak stresses are up to 2 times higher than ultimate peak T and C stresses for human cortical bone (T: 120-150 MPa, C: ~166 MPa) (Whiting & Zernicke, 2008, 2<sup>nd</sup> ed). However, higher strain rates that occur during impact activities, like running, lead to bone withstanding larger ultimate stresses than typically published (McElhaney, 1966). For example, at strain rates of 0.001 Hz, ultimate compressive stress for the femur is 150 MPa, but it increases to 317 MPa at 1500 Hz. So at physiological strain rates observed during running of approximately 0.2-0.3 Hz (Altman & Davis, 2012), ultimate compressive strain is between 1.33-1.47 times larger than "static" (i.e., 0.001 Hz) testing values, which would be 200-221 MPa, for the femur sample tested (McElhaney, 1996). Additional limitations of the study include the novel conditions only representing acute changes. Runners may make further adaptations after regularly using a new foot strike style/running BF (Warne & Warrington, 2013).

## Conclusion

Peak T, C, and S tibial stresses are greatest during shod mid/forefoot striking and lowest for shod rearfoot striking (except for peak T being lowest during BF for hRF). Stresses were generally intermediate for BF. Therefore, it may not be advisable for hRF to switch to a shod mid/forefoot strike due to the higher peak tibial stresses. But for those runners choosing to switch, they should do so gradually to allow adequate time for bones, and other tissues, to adapt to the higher stresses to help avoid injury. An 8% shorter stride length only significantly decreased S stresses, but not T or C stresses. A larger change in stride length (e.g., 10% decrease) may be necessary to significantly decrease stress fracture risk.

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## CHAPTER 8

### SUMMARY AND CONCLUSIONS

#### Summary

In my first study, we found that impacts exist in both RFS and FFS running. With a RFS, the impact is evident in the vertical ground reaction force, but with FFS, impacts are present in the posterior and medial directions. The loading rates are higher in the posterior and medial directions for FFS and higher in the vertical direction for RFS, in general. Habitual rearfoot strikers were able to decrease their vertical GRF variables by using a FFS. However, habitual mid/forefoot strikers did not necessarily have lower values when using a FFS compared to a RFS.

From my second study, in general, joint moments and contact forces were larger at the ankle for FFS while they were larger at the hip for RFS or not different between foot strike styles. Some knee moments and contact forces were larger for RFS while others were larger for FFS. Patellofemoral loading was similar for foot strike styles. Nearly all moments and contact forces decreased when runners used a shorter SL, and a majority of the variables that were elevated for RFS compared to FFS were successfully reduced by using a shorter SL. Additionally, we found that the free moment was distinctly different between the two foot strike styles. Step width was narrower for FFS, but ITB strain and strain rate were similar between foot strike styles. RFS was associated with greater pelvic drop. Shortening one's SL had many minor beneficial effects, including a wider SW and decreased free moment, pelvic drop, hip adduction, ankle eversion, ITB strain, and ITB strain rate.

In my third study, stride interval long range correlations became more patterned during BF running and using a novel foot strike style.

In my final study, shod FFS resulted in higher compressive, tensile, and shear stresses in the distal tibia compared to shod RFS and BF. Only shear stresses significantly decreased with an 8% shorter SL.

## Conclusions

In conclusion, of the variables we have studied, external loading (GRF and LR) is larger in the different directions for shod RFS and FFS. Internal loading (moments, contact forces, bone stresses) is higher at different joints and in different planes for RFS and FFS, although tibial stresses appear to be larger for shod FFS compared to RFS and BF. Additionally, using a novel foot strike may lead to less adaptability. Therefore, one foot strike style does not ubiquitously decrease loads so it alone will not decrease risk of all overuse injuries. A shorter SL, alternatively, decreases lower extremity moments, contact forces, and some kinematics related to injury. Tibial stresses may not significantly decrease with an 8% shorter SL, though.

Future studies will continue to explore different variables related to this controversial topic. Studies that would significantly advance this area of research include: 1) large study of running injuries most commonly associated with the different footwear and foot strike styles, 2) prospective study of biomechanical changes and running injury prevalence in habitual rearfoot strikers who switch to a mid/forefoot strike and/or to minimalist shoes or barefoot running, 3) prospective study of changes in bones properties of several lower extremity bones during such transitions, and 4) external validity and long-term retention of shortening runners' stride length outside the laboratory.

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